Advances in Medicine and Engineering Interdisciplinary Research, 2026, Volume 4, Issue 3
DOI: 10.32629/ameir.v4i3.5470
ISSN Online: 2972-3825
ISSN Print: 2972-3833

Denoising of magnetocardiography signal and calculation of two-dimensional magnetic field mapping

Wenhao Li*

Beijing Haidian Daoxiang Lake School, Beijing, China
*Corresponding author.
Email address: h2027198@thisdl.edu.cn

Abstract: Magnetocardiography (MCG) is a non-invasive cardiac detection technology that is radiation-free and is suitable for all populations. But the raw magnetocardiography signal is polluted by power noise, baseline drift, and pulse artifacts, thereby reducing the quality of cardiac magnetic imaging and electrophysiological analysis. This paper utilizes an integrated denoising method and gives two-dimensional magnetic field mapping calculation process. Serial multi-stage denoising chain, including bandpass filtering, notch filtering, median filtering, and wavelet filtering is applied to suppress various noises, while retaining the magnetocardiography waveform characteristic. And gives the calculation of isomagnetic map, pseudo current density map, and curl map.

Key words: magnetocardiography; denoising; isomagnetic map; pseudo current density map; curl map

1 Introduction

Magnetocardiography (MCG) is an emerging biomagnetic measurement method. It detects the magnetic field of the heart by shielding the earth magnetic field [1]. Cardiac magnetic detection is non-invasive. Unlike some other detection methods (such as PET, SPECT, etc.), cardiac magnetic detection is suitable for all people, including children and pregnant women. Cardiac magnetic detection can not only be used to detect existing diseases, but also to identify potential diseases in the human body [2,3]. When the magnetocardiography equipment generates signals, it will be disturbed by the surrounding environment and thus produce signal with noise. Therefore, it is necessary for us to preprocess the magnetocardiography signals to obtain a signal that mainly contains the signal of the heart [1]. Wavelet and EMD-type decomposition can layer-by-layer strip power frequency and electromyographic interference. The improved variational mode decomposition can adaptively separate the low-frequency baseline drift, and precisely filter out the noise of each mode by combining the interval threshold. Independent component analysis is adapted to multi-channel cardiograms to separate environmental magnetic interference [1]. All kinds of adaptive filters can track drift noise in real time. Deep learning and optimizing the hybrid decomposition scheme can significantly enhance the output signal-to-noise ratio, retain core waveforms such as QRS and ST, and also generate isomagnetic maps and pseudo-current density imaging based on the processing results to assist in the interpretation of cardiac lesions. In this paper, a composite magnetocardiography signal denoising method is adopted, and calculations are carried out on the isomagnetic map, pseudo current density map and curl map.

2 Magnetocardiography signal denoising

In dealing with the noise of magnetocardiography signals, this paper employs a variety of filters. It includes bandpass filtering, notch filtering, median filtering, and wavelet filtering. The entire magnetocardiography signal denoising process is carried out in a serial manner in the pre-determined sequence. Firstly, the effective signal frequency band is defined by relying on the 0.5-40 Hz bandpass filter, and the ultra-low frequency baseline drift and high-frequency irrelevant noise are eliminated to complete the initial regularization of the signal frequency band. By leveraging the 50 Hz notch filter, the narrowband strong interference of the power grid's power frequency, which is highly targeted, is eliminated to avoid the interference caused by this strong noise to the subsequent time-domain filtering. Then, the 15-point median filter is relied on to filter out the burst impulse glitch noise and clean up the abnormal jumps in the time domain. Subsequently, the remaining gentle random noise is finely weakened through wavelet threshold noise reduction, and the clean base after the previous layer noise reduction is relied on to avoid the false elimination of the cardio magnetic characteristic waveforms. Each step performs its own function, and a step-by-step purification noise reduction chain is established based on the preprocessing results of the previous step. The previous step creates stable and high-quality signal input conditions for the subsequent step, gradually stripping away different types of noise. The usage of filtering methods can be classified as the following steps.

First of all, the principle of bandpass filtering [4] is to only allow signals within a specific frequency range to pass through, while filtering out low-frequency and high-frequency components that are below the lower limit cut-off frequency and above the upper limit cut-off frequency [1]. Suppressing both low and high frequencies while retaining only the passband frequency in the middle is used to extract specific frequency band features.

Supposing that f_L denotes the lower cut-off frequency, f_H denotes the upper cut-off frequency, $H_{BP}(f)$ denotes the frequency response of bandpass filter, then:

$$H_{BP}(f) = \begin{cases} 1, f_L \le |f| \le f_H \\ 0, \text{other} \end{cases}$$

In this paper, f_L=0.5 Hz, f_H=40 Hz.

Second, notch filtering [5] belongs to linear frequency-domain filtering and is of the same type as bandpass, but its functions are completely opposite: bandpass retains only one frequency segment and filters out the rest. Notch waves retain almost all frequencies and precisely eliminate only one extremely narrow interference frequency point. It is specifically designed to handle single fixed-frequency interference, such as 50 Hz industrial frequency AC noise, fixed harmonics of motors, and inherent vibration noise of equipment in industrial collected signals. This type of interference is concentrated at only one frequency. Notch waves only cut out this one frequency point, while the rest of the signal is completely retained. Supposing that f_0 represents the notch frequency that needs to be suppressed, and Δf is the minimum bandwidth of the stopband, then

$$H_{Notch}(f) = \begin{cases} 0, |f - f_0| \le \Delta f \text{ or } |f + f_0| \le \Delta f \\ 1, \text{other} \end{cases}$$

where $H_{Notch}(f)$ denotes the frequency response of notch filter. In this paper, f_0=50 Hz, Δf=1 Hz.

Thirdly, median filtering [6], which does not belong to frequency-domain processing, is a nonlinear filtering method that directly performs local window operations in the original signal. Its core objective is to remove pulse-type and isolated sudden noise. Median filtering takes advantage of the mathematical property that the median is not affected by extreme mutation values to eliminate isolated noise while preserving edges and waveform mutation positions. Its one-dimensional discrete mathematics description is:

Letting the sliding window length be 2k+1, for the sampling point x[n], the window sequence is:

$$W[n] = \{x[n-k], x[n-k+1], \ldots, x[n], \ldots, x[n+k]\}$$

Sort the data within the window in ascending order:

$$W[1] \leq W[2] \leq \ldots \leq W[2k+1]$$

Output median:

$y[n]=\text{Median}(W[n])=W[k+1]$

In this paper, k=7.

Fourth, wavelet filtering [7]. The traditional Fourier transform can only look at the global frequency and cannot locate at which time or local position of the image the noise appears. Wavelet transform realizes dual localization analysis of time and frequency, decomposes the signal into components of different scales in layers, distinguishes the signal from the noise in layers, and filters out the noise in layers. It is one of the optimal methods for handling non-stationary signals. In this paper, db4 wavelet base is used, and the decomposition level is 5. Figure 1 shows the magnetocardiography signal of one channel after denoising.

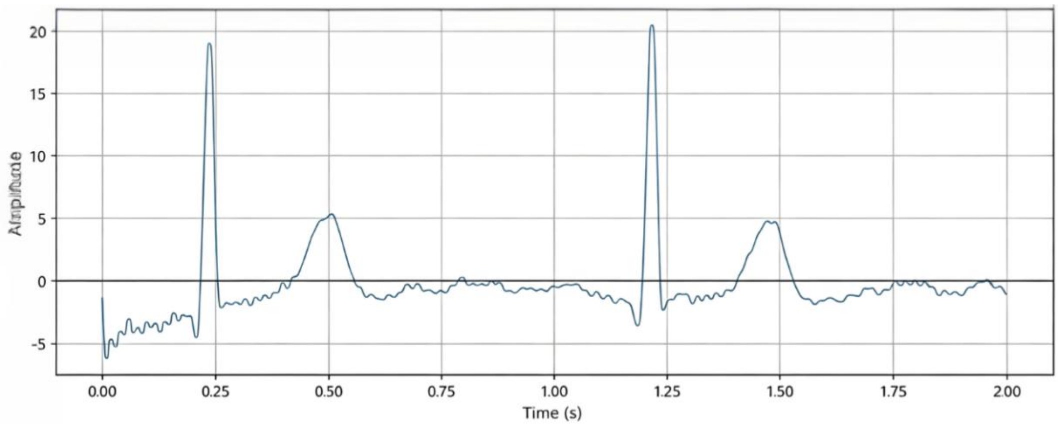


Figure 1. The magnetocardiography signal after denoising

3 Two-dimensional magnetic field mapping calculation

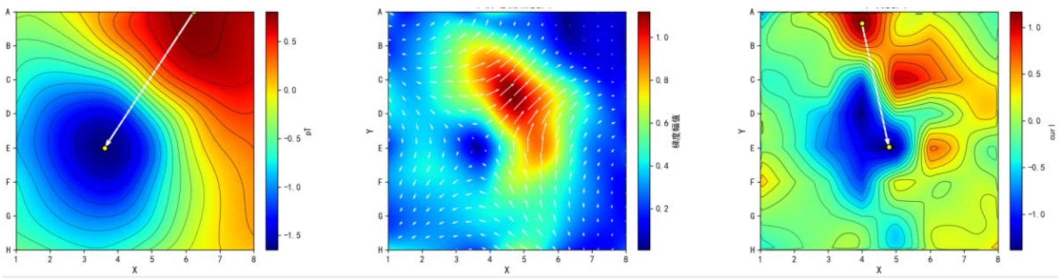
This process implements a complete spatial computational method for magnetocardiographic signal analysis, transforming discrete multi-channel sensor data into, magnetic field maps, pseudo current density distributions, and vortex field characteristics. The entire computation begins with one-dimensional channel amplitude vectors collected by a square array of magnetic sensors, where each vector represents the magnetic field strength at a given sampling time for each sensor channel.

To address the issue of low spatial resolution caused by sparse sensor placement, a bicubic spline interpolation method is employed to densify the original low-density sensor grid into a fine, uniform mesh. This interpolation technique was selected due to its superior performance in spatial field reconstruction, as it preserves smooth, continuous surface features and avoids abrupt numerical variations. Unlike simple linear interpolation, bicubic spline interpolation [8] achieves high-precision fitting of spatial magnetic field variations through local data approximation, effectively eliminating discontinuities between sensor sampling points. Based on detailed isomagnetic maps, this paper further calculates the pseudo current density field, which is the core physical parameter for interpreting cardiac electrical activity. According to quasi-static electromagnetic theory applicable to biomagnetic signals, the equivalent current distribution [9] within a plane can be reconstructed from the first-order spatial gradient of the normal magnetic field component. The horizontal and vertical spatial gradients of interpolated magnetic field maps is computed using numerical differentiation, then constructs two orthogonal components of the pseudo current field based on established rules for solving biomagnetic inverse problems. By combining these two vector components, the magnitude of the pseudo current density is determined, forming a two-dimensional contour map that characterizes the intensity and distribution patterns of equivalent cardiac currents the formula is as follows:

$$\begin{cases} J_x = -\dfrac{\partial B_z}{\partial y} \\ J_y = \dfrac{\partial B_z}{\partial x} \end{cases}$$

where J_x denotes current density in the x-direction, J_y denotes current density in the y-direction, B_z denotes magnetic field in the z-direction.

Combined with vector arrow visualization techniques, this approach clearly reveals the direction and strength of cardiac current flow. Based on the current density field, this paper calculates the spatial curl of the current field to characterize the vortex features of cardiac electrical activity. The curl parameter reflects local rotational changes within the planar current field, corresponding to the second-order spatial variation of the original magnetic field distribution. By solving partial derivatives of current vector components along different spatial directions and computing their differential combinations, a curl distribution map is generated. This can effectively identify regional convergence and vortex activities in myocardial currents [9], serving as a crucial indicator for analyzing abnormal cardiac electrical excitation. Figure 2 shows the results of isomagnetic map (a), pseudo current density map (b) and curl map (c) at 270ms.



(a) (b) (c)

Figure 2. Isomagnetic map, pseudo current density map and curl map

4 Summary

Cardiac magnetic imaging is a non-radiative and universal cardiac examination technique used to detect both overt and recessive heart diseases. However, the raw MCG signal is affected by power frequency interference, baseline drift, impulse noise, which hinders imaging analysis. This paper realizes one-dimensional MCG signal denoising and two-dimensional magnetic field mapping. A serial multi-step noise reduction chain combining 0.5-40Hz bandpass filtering, 50 Hz notch filtering, 15-point median filtering, and wavelet filtering is constructed to remove multiple types of noise while maintaining waveform characteristics. The isomagnetic map, pseudo-current density map, and curl map are calculated.

Conflicts of Interest

The author declares no conflicts of interest regarding the publication of this paper.

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