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Selective Averaging of QRS Complexes in Magnetocardiograms

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Abstract. Averaging of QRS complexes is often used to improve the signal to noise ratio in magnetocardiography (MCG). Simple trigger based averaging procedures of QRS complexes, however, ignore the variation in amplitude and shape of the signals caused e.g. by respiration. A suppression of signal portions within the QRS complexes may occur. Additionally, for inverse source reconstructions of dipoles and of current density distributions errors in the spacial arrangement may occur.
In order to overcome these problems we developed a method for separating and selective averaging QRS complexes with different shapes and amplitudes. The method is based on a spline interpolation of the QRS complex averaged by a standard procedure. Then, this spline function is fitted to each QRS complex in the raw data by means of the Levenberg-Marquardt method. Five regression parameters are applied: a linear amplitude scaling, two parameters describing the baseline drift, a time scaling parameter, and a time shift parameter.
Both amplitude and shape of the QRS complex are influenced by respiration, while the baseline shows a weaker influence of the respiration. We found a linear correlation of the regression parameters of two neighboring measurement channels. Therefore, selective averaging of a larger number of sensors can be performed simultaneously. However, the respiration caused QRS amplitude variability is not directly correlated to the absolute QRS amplitude in multichannel MCG.
In conclusion, we believe that selective averaging of QRS complexes will improve signal analysis and source reconstructions.

Keywords: *Magnetocardiography, MCG, QRS complex, selective averaging, spline interpolation*

Introduction

Electrocardiograms (ECGs) and magnetocardiograms (MCGs) are mainly used in two ways: (i) signal analysis in time and frequency domain to extract relevant features such as e.g. late potential parameters [1,2,3,4,5] and (ii) source reconstructions to noninvasively localize focal or distributed sources in the heart [6,7,8]. Signal averaging is often used to improve signal to

noise ratio (SNR) [7]. The averaging procedure most commonly applied consists in finding a trigger list of maximum correlation points by correlating a chosen QRS template to the raw signal, and then averaging the QRS complexes according to the trigger list found. This method is also called maximum coherence matching (MCM) with a template beat [9]. The disadvantage of the MCM method is its low selectivity, i.e. QRS complexes of various amplitudes and shapes are averaged.
Two main causes for beat to beat variability can be distinguished. The first group of causes consist in patient movement artefacts and respiration in particular which alter the position of the sources relative to the measurement channels and also alter the surrounding volume conductor. Thus, source reconstructions suffer from two additional errors: the displacement of the sources and the volume conductor relative to the sensors and the changes within the volume conductor. Simulations show that volume currents and thus a change in the extension and position of the lungs influence the measured magnetic field [10]. Most of these problems can be overcome by selectively averaging QRS complexes, which is proposed in this paper. The second group of causes for beat to beat variability, which are not considered in this abstract, are due to the pathology in patients with an electrically instable myocardium.

In a previous paper [11] we discussed QRS amplitude and shape variability in magnetocardiograms which were mainly caused by respiration. In this abstract we review these results in regard to selective averaging of QRS complexes.

Methods

Measurements

The magnetocardiagram of a healthy volunteer was measured in a magnetically shielded room (Vacuumschmelze Hanau, AK3b) at the Biomagnetic Center Jena using the 62 channel biomagnetometer (Philips, Hamburg) [12]. The healthy state of this volunteer was proven by a history free of cardiac symptoms, normal physical examination, normal 12-lead electrocardiogram, and normal findings in echocardiography. The subject was lying in a supine position and the two dewars were positioned above the thorax so that they covered the field maxima. The subject was instructed to breathe normal but very uniform. We selected the channel providing the highest signal amplitude at R peak. The position of the pick-up coil of this channel was approximately 12 mm right and 170 mm below the jugulum. We recorded a time period of 100 s at a sampling rate of 1000 Hz. An analog first order highpass filter (0.036 Hz, 3 dB) was applied to the analog signals. An example of two QRS complexes measured is given in Figure 1.

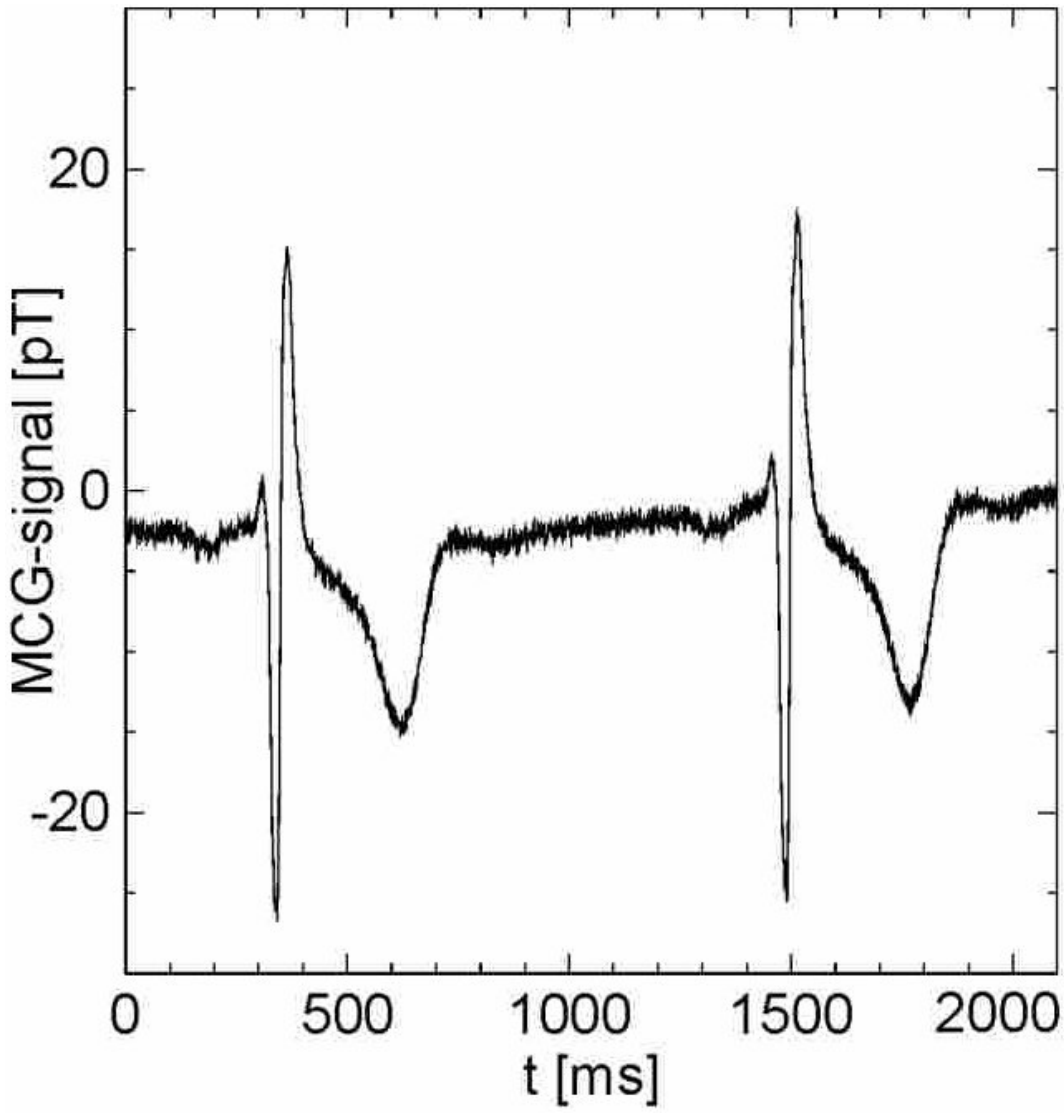

Figure 1. Example of a measured MCG-signal.

Modeling and Computation

A scalable QRS function SQRS is used to model the QRS complexes:

$$S_{QRS}(t) = A \cdot N_{QRS}\,(L \cdot t - t_{beat}) + S_0 + S_1 \cdot (t - t_{beat})$$

The function N_{QRS} (t) is the normalized QRS signal. We use the parameters A (amplitude), t_{beat} (event time), and (S_0 , S_1) (describing the linear baseline) for every QRS complex. The parameter L describes changes in the length of QRS complex (shortened and prolonged QRS complexes).

We use a standard MCM method to compute a first average of the QRS complex. Next, the averaged QRS complex is used as a new template for a second MCM computation. The normalized signal N_{QRS} (t) is then estimated by performing a third order spline interpolation on the averaged QRS signal calculated with the two step MCM method. Figure 2 shows the averaged signal of a one heart cycle and the spline interpolation of the averaged QRS complex.

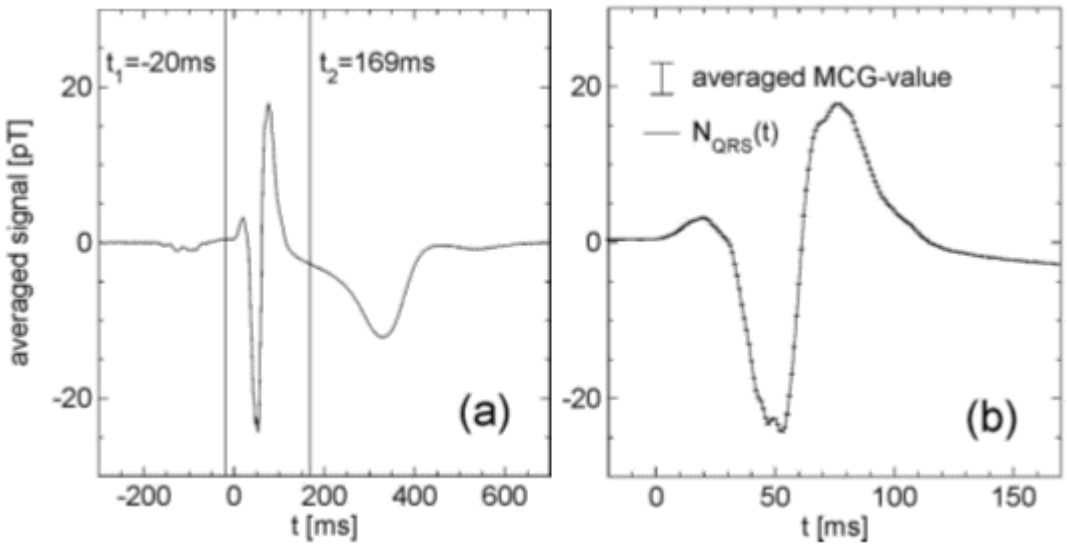

Figure 2. Averaged MCG-signal with QRS starting at t=0 ms and spline interpolation of the averaged signal in the time interval from -20 ms to 169 ms.

Subsequently, the spline function $N_{QRS}(t)$ is employed to fit the function

$$S_{QRS}(t) = S_{QRS}(t, A, S_0, S_1, t_{beat}, N_{QRS}(t))$$

to every QRS complex through a nonlinear regression (Levenberg-Marquardt-Method [13]). The regression parameters are the amplitude A, the time event t_{beat}, and linear baseline parameters (S_0, S_1). In order to avoid spline interpolation errors introduced by the fitting interval edges, we use a shorter time interval (10 ms on both sides) for the nonlinear regression. The regression for L is applied in the time interval 30 ms to 89 ms (QRS complex duration), where this parameter is relevant.

Results

Figure 3 shows the estimated regression parameters A, t_0, S_0 , S_1 and L versus the beat time t_{beat} for all QRS complexes. The errors (bars) are calculated from the adequate regression covariance matrix.

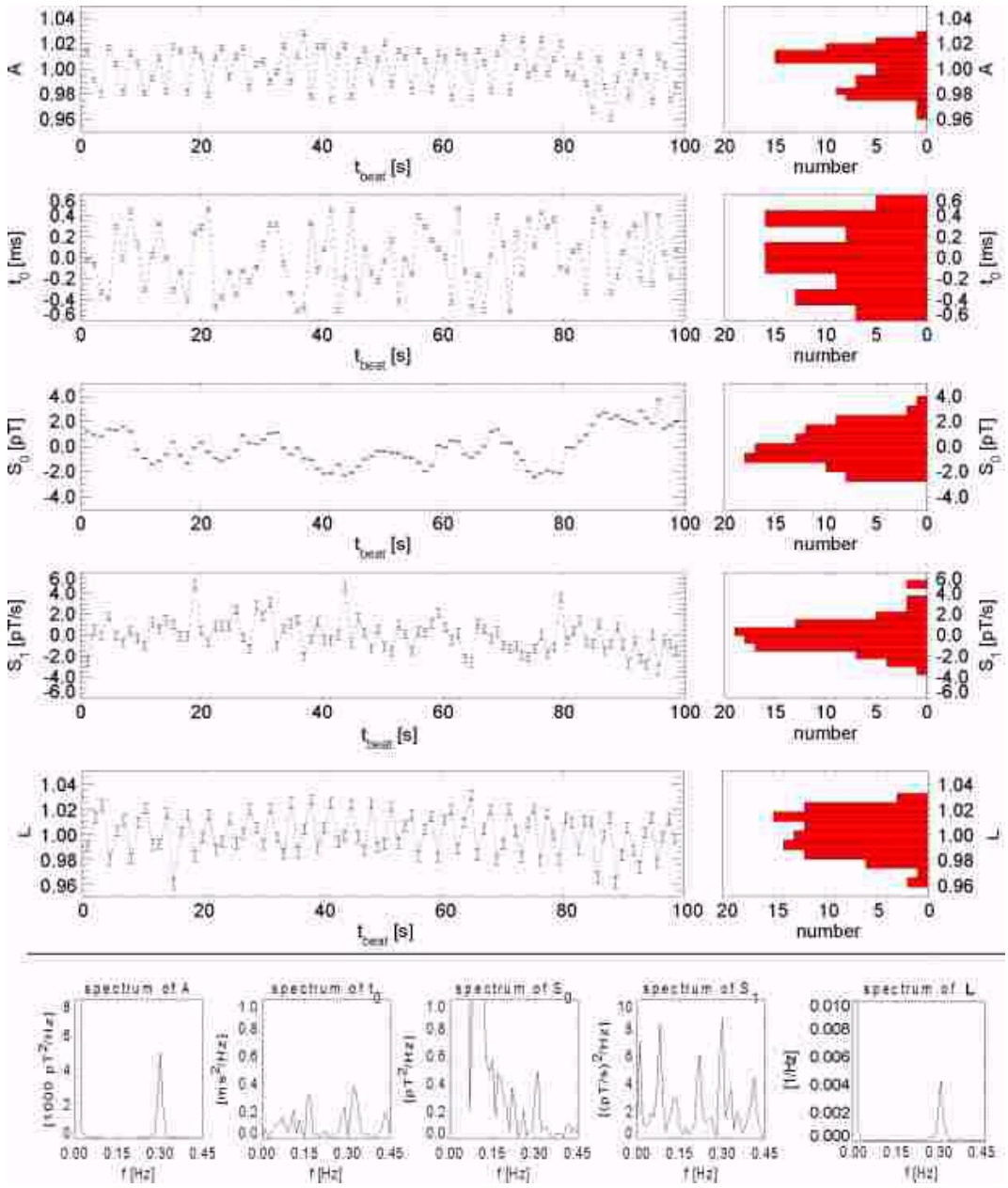

Figure 3. Regression parameters over t_{beat}, their empirical distributions, and power spectra.

The modulation of both *A* and *L* fluctuate up to ± 3 % and have a typical cycle duration of the respiration. The histograms on the right side in Figure 3 illustrate that the distribution basically consists of two peaks, the inhaled and exhaled state peak, respectively. The power spectrum of the amplitude in Figure 3 exhibit a peak at approximately 0.30 Hz, which equals 18 respiration cycles. The time series of the linear baseline parameters (S_0, S_1) shows an influence of environmental noise on the linear baseline. However, the power spectra have a small peak at 0.30 Hz, that means a weak influence of respiration on the linear baseline.

There was no significant cross correlation between all parameters.
MCG based source reconstructions require the use of multi-channel systems. During the MCG recording it is important that the arrangement of signal sources and the volume conductor is fixed with respect to the sensor positions. Therefore, we quantify the influence of the alternating arrangement caused by respiration on the signal preprocessing in sensor arrays. As a first step we analyze the cross correlation of the estimated regression parameters for two adjacent pick-up coil positions.

In Figure 4 the regression parameters of both sensors for 90 heart cycles are plotted against each other, where the statistical regression errors are shown as bars. The solid lines illustrate the linear regressions. The gradient of the linear regression function is 1.5 for the amplitude *A*. This means, respiration causes a greater absolute fluctuation of the QRS amplitude at the position of sensor 2 than at the position of sensor 1 although the absolute signal is larger in sensor 1.

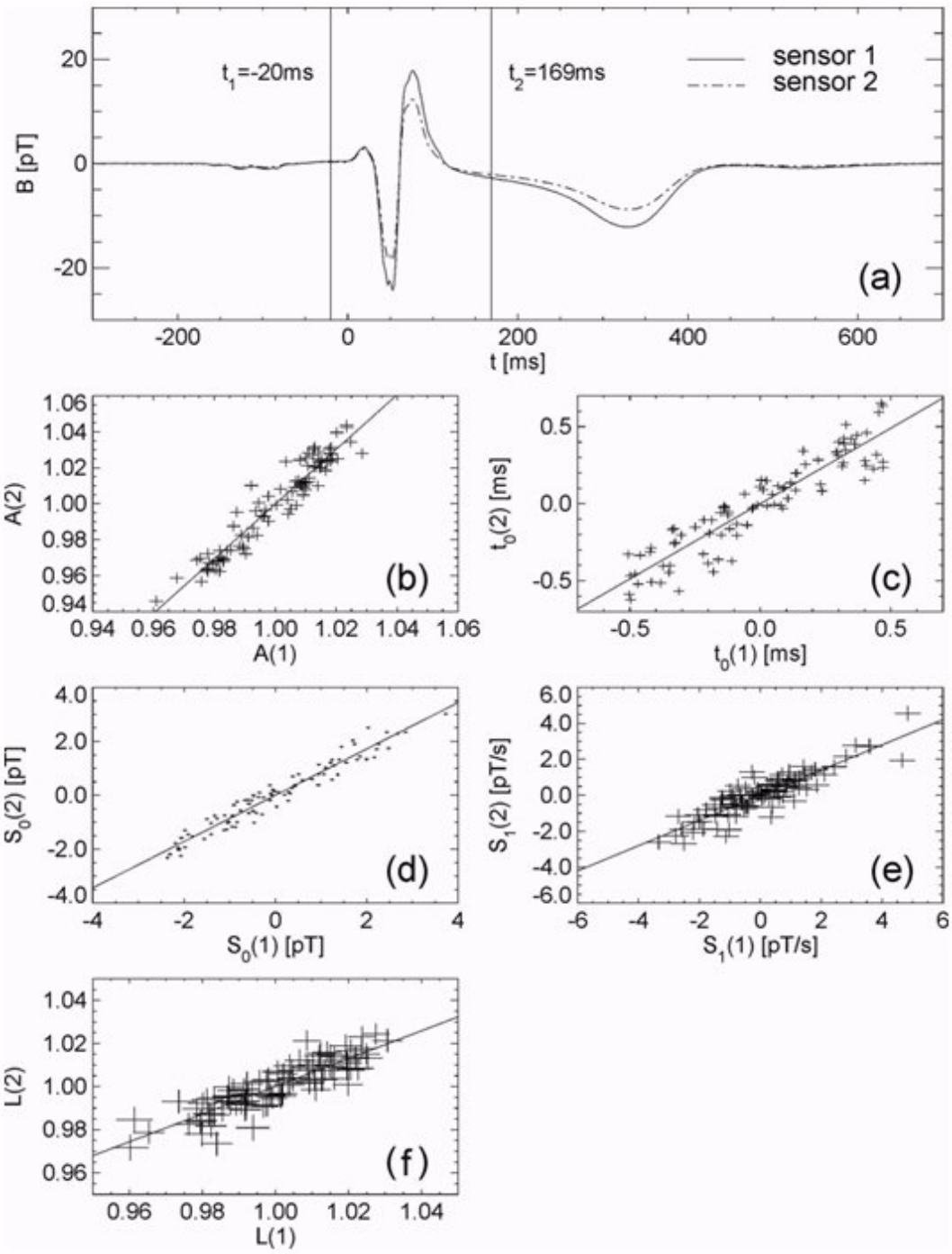

Figure 4. Averaged signals (a) and linear cross correlation for the regression parameters of two adjacent measurement channels (b-f) at the MCG field maximum.

Conclusions

The amplitude and the shape of the QRS complex are influenced by respiration. Moreover, respiration can produce nonlinear changes in the field strength of multi channel MCGs. This produces corrupted results in source reconstructions. Selective processing and averaging of heart cycles for the same phase of respiration can reduce such a corruption. If source reconstructions are constrained by anatomical information (e.g. the source space is restricted to the heart surface) this information is to correspond to the respiration cycle. This information can be obtained by using a respiration trigger in the MRI scans. The regression parameters of two adjacent measurement channels correlate linearly. Thus, selective averaging of a larger number of sensors can be performed simultaneously.

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