Abstract – A medical application of ultra-wideband radar is proposed where it is used for monitoring motion of internal organs, especially due to respiration and heart beat. This provides complementary information, e.g. for improved cardiac magnetic resonance imaging. To this end, a blind source separation is applied to decompose the multitude of physiological signatures after selecting a region of interest in the received UWB-data. By extending our radar system to multiple channels we have established more favorable conditions for the solution of this ill-conditioned problem. The diagnostic potential of the cardiac motion component by UWB-radar is shown by a time-domain comparison with corresponding electrocardiograms and in a setup where the heart’s left and the right functional parts were illuminated simultaneously.

1 INTRODUCTION

Due to the recent developments in ultra-wideband (UWB) technology, this versatile technique gains more and more recognition in essential sensing and diagnostic applications. The specific advantages, like non-contact sensing and high temporospatial resolution, good tissue penetration, low integral power, and compatibility with established narrowband systems, favour UWB radar for biomedical applications. Our primary aim is the technological development of UWB radar combined with magnetic resonance imaging (MRI) to support innovative fields such as cardiac MRI (CMRI) by a UWB navigation technique [1,2]. Up to a magnetic field of 1.5 T an electrocardiogram (ECG) signal is routinely used for gating cardiac MRI. However, due to the magneto-hydrodynamic effect, ECG triggering is seriously hampered in high and ultra high-field MRI systems and alternative gating techniques are needed.

The basic principle of motion detection by UWB radar is to determine the displacement of tissue interfaces by measuring the propagation time of the reflected electromagnetic waves originating from these interfaces. However, the recorded UWB raw data contain a mixture of all simultaneously occurring physiological movements. To extract the CMRI relevant information from this composite signal, the reliable decomposition of the individual geometrical displacements is mandatory. Thus, the MRI data have to be unambiguously assigned to the phases of both, respiratory and cardiac cycle. To this end, we have selected a region of interest in the impulse response functions (IRF) of the raw data to construct artificial signal channels, which are subsequently processed with a blind source separation (BSS) approach, aiming at decomposing the multitude of physiological signatures. For comparative reasons, we analyzed the motion components obtained by UWB together with simultaneously acquired ECG data.

2 MATERIALS AND METHODS

Our proposed method consists of three steps: 1. the UWB measurement, 2. signal-source decomposition by BSS, and 3. signal processing to extract cardiac events.

In a first study, we measured UWB and ECG simultaneously and performed the described data processing steps. Secondly, we measured the cardiac motion by UWB radar with two separate groups of antennas, simultaneously and independently illuminating the left and right sides of the thorax without any ECG recording.

2.1 Combined UWB-ECG measurement

UWB and ECG were simultaneously acquired using a radar system with one transmitter (Tx) and two receiver (Rx) channels. The Tx and one Rx antenna were facing the antero-posterior direction, the second Rx antenna was oriented towards the left-anterior oblique direction (Fig. 1). The ECG was recorded with two channels (left arm and left leg against right arm). UWB and ECG data were recorded at 44.2 Hz and 8 kHz, respectively.
Figure 1: Setup for simultaneous UWB and ECG measurements.

The transmitted radar signals were generated by a pseudo-random M-sequence with a length of 511 clock signals at \( f_c = 8.95 \) GHz. The equivalent UWB power spectrum extends up to \( f_c/2 \). A lower frequency limit is given by the cut-off frequency of the antennas at 1.5 GHz [2]. The IRF is obtained by correlating the received signal with the M-sequence [3]. In this way, we obtain IRFs with a length of 511 data points per sample.

Figure 2: Radargram with selected data points

Every data point in the IRF can be considered the response of the object depending on the distance to the antennas. In the IRFs of all scans, 100 time points were selected for the region of interest (ROI, see Fig. 2) and transformed to 100 artificial data channels. The selected interval starts after the IRF maximum, which results from the direct crosstalk between Tx and Rx antennas. In this manner, 200 artificial data channels are obtained from the IRFs of two real UWB channels and are available for decomposition by BSS.

2.2 UWB measurement with multi-illumination

In the second measurement configuration (Fig. 3), we applied two groups of four Rx and one Tx channel in left lateral and right anterior oblique positions.

Figure 3: Setup with two antenna groups, one on the left lateral side and one on the right anterior oblique.

For this purpose, a radar device (8 Rx, 4 TX) with a sampling frequency of 132.6 Hz was employed. Each antenna group consists of one Tx antenna surrounded by four Rx antennas. All antennas were pointed to the estimated center position of the heart. Similarly as described before, this gives 400 artificial channels per group with four real Rx channels.

2.3 Signal source decomposition by TDSEP (BSS)

The blind source separation assumes a measured signal \( x(t) \) to be a linear combination of unknown zero-mean source signals \( s(t) \) with an unknown mixing matrix \( A \):

\[
x(t) = As(t) \quad x = (x_1, ..., x_m)^T
\]

The components \( y(t) \) estimating the original sources \( s(t) \) can be calculated from the estimation of the de-mixing matrix \( A^* \approx A^{-1} \):

\[
y(t) = A^*x(t) = A^*As(t) = LPs(t)
\]

The components \( y(t) \) estimate the original sources \( s(t) \) up to a scaling factor \( L \) and a permutation \( P \). BSS algorithms usually assume statistical independence of the sources \( s_i(t) \), i.e. the joint probability density function of the signals factorizes. The statistical independence can be tested by calculating the time-shifted cross-correlation between two sources \( s_i \) and \( s_j \). However, under the assumption of non-zero auto-correlations, the time delayed correlation matrices have to result in diagonal matrices for all time shifts \( \tau > 0 \). We applied a second-order time-domain algorithm (TDSEP, Temporal Decorrelation source SEPa ration) [4, 5]. In TDSEP the unknown mixing matrix
A is calculated by simultaneous diagonalization of a set of correlation matrices \( R_{\tau}(x) \) for different choices of \( \tau \).

\[
R_{\tau(x)} = \langle x(t)x^T(t-\tau) \rangle,
\]

\[
R_{\tau(x)} = \left( A_s(t)(A_s(t-\tau))^T \right) = \mathbf{A} R_{\tau(x)} \mathbf{A}^T,
\]

where the angular brackets denote time averaging. The quality of the signal separation depends strongly on the choice of \( \tau \). However, solving \( R_{\tau(x)} = \mathbf{A} R_{\tau(x)} \mathbf{A}^T \) for several \( \tau \) by simultaneous diagonalization eliminates this obstacle. For biomagnetic applications, typically more than 40 time shifts \( \tau \) should be chosen and time constants of those components which are known \textit{a priori}, e.g. the possible cardiac frequencies \( 1/\tau_{\text{cardiac}} \), should be included. Additionally, Principal-Component Analysis (PCA) compression was applied to reduce the number of channels used for generating the correlation matrices \( R_{\tau(x)} \) and reduce computation time for the BSS. The components of the resulting sources are calculated via Eq. (2).

Automatic identification of the cardiac component was provided by a frequency-domain selection criterion. The algorithm searches for the highest ratio \( I_{\text{in}}/I_{\text{out}} \), where \( I_{\text{in}} \) is a single narrowband signal (fundamental mode and first harmonic) within the frequency range of 0.5 Hz to 7 Hz and \( I_{\text{out}} \) is the maximum signal outside this range. This criterion utilizes the fact that for non-pathological conditions the main spectral power density of the heart motion falls in this frequency range.

2.4 Signal processing to extract cardiac events

To the cardiac component of the UWB signal a high-order zero-phase digital band-pass filtering of 0.5–5 Hz was applied. In datasets acquired under continuous breathing, the BSS component with the highest power reflects respiration and this component can be identified via the highest L2–norm. For datasets with non-continuous breathing it is more reliable, however, to apply a band-pass filtering of 0.05 Hz to 0.5 Hz to the sign dependent sum of the DC-free UWB signals. Prior to their comparison, the cardiac UWB and ECG signals were both re-sampled at 1 kHz to retain the more detailed information of the ECG. To trigger on the latter signal, the usual R-peak detection was applied. In the UWB signal, we chose the points of maximum myocardial contraction during the cardiac cycle, i.e. the minimum of the radar cross section. These points appear as minima in the UWB signal (Fig. 4: red crosses). To enhance the robustness of this detection scheme, we combined it with a simple difference calculation at the trailing edge of the minima.

3 RESULTS

In the comparison of the cardiac UWB component and ECG, we have to consider the cardiac mechanics as well as the cardiac electric activity. Related to the R-peak, indicating the point of the myocardium’s maximal electrical activity, the point of maximal mechanical contraction is delayed. More important, however, for MRI gating is the existence of a rigid relation between trigger points selected by ECG or UWB. The delays between ECG and UWB trigger events varied between 392 ms and 468 ms, with a standard deviation of 18.3 ms. This is already smaller than the UWB sampling time of 22.62 ms and can be considered an excellent result, proving the consistency of our procedure.

We validated the proposed triggering scheme by measurements of non-uniform respiration and partial
breath holding. An example for the latter is shown in Fig. 5, where the algorithm rejected several false trigger events correctly. One real heart beat, on the other hand, at \( t = 45.5 \) s, was rejected, too. This particular beat exhibits quite a singular R-R-duration in the ECG, however, and we can assume a different mechanical contraction compared to the other beats. From the CMRI point of view, it is presumably even beneficial to have it excluded as image quality suffers from such arrhythmic cycles.

Figure 6: cardiac motion with a) one, b) two and c) four Rx channels

In the case of double-sided illumination, especially in the left lateral position, the attenuation of the reflected signals from the heart is much higher. This is due to the prolonged propagation path through the conducting tissue. Consequently, the UWB signal from the cardiac motion is weaker and much more affected by noise, Fig. 6 a). The signal quality improves considerably, however, when the number of Rx channels is increased, as shown in Fig. 6 c). We have examined only healthy volunteers up to now, but even there the diagnostic potential is recognizable at the differences in velocity of contraction (trailing edge) and dilatation (rising edge) and by detection of differences in time, like the prolonged duration of the dilatation on the right ventricle. In the next step the application to pathological pattern is needed.

4 DISCUSSION

With the proposed method, we are able to reliably determine trigger points in the cardiac UWB signal. Physiologically correct, these triggers are lagging the corresponding R-peak of the ECG by a fixed delay of about 400 ms, which is a certain disadvantage for CMRI applications. By monitoring cardiac mechanics rather than electric functionality, the proposed non-contact method is even better suited, on the other hand, to provide distortion-free, high-resolution medical imaging. Even the detection of spontaneous changes in the cardiac mechanics (extra systoles) seems to be possible. The analysis of cardiac mechanics accessible by stand-alone UWB-Radar or in conjunction with the electrical activity from the ECG contains beneficial diagnostic information, e.g. for infarction detection, as ischemic tissue shows a modified contraction pattern.

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