

# Skip the Electrodes, But Not A Beat: The Engineering Behind the Beat Sensor

Peter Speier, Ph.D. (manuscript) and Mario Bacher (box texts)

Siemens Healthineers, Erlangen, Germany

## Introduction

Early in the development of cardiac MR, even long before bSSFP/TrueFISP made it clinically useful [1], the ECG was established as the standard trigger source [2] despite its well-known drawbacks of complex patient preparation and MR-related artifacts caused by the magnetohydrodynamic (MHD) effect and the gradient pulses of the MR measurement [3, 4]. Its main advantage is the early and well defined trigger time point based on the R-wave that precedes the ventricular contraction.

The only other clinically adopted cardiac trigger method is the finger-tip pulse sensor peripheral plethysmogram (PPG) [5], but its late and patient-state-dependent trigger position in the cardiac cycle reduces both its versatility and stability. Therefore, PPG is used only as a back-up.

Over the years, the reliability and ease of use of the ECG have been improved dramatically by the introduction of the vector ECG [6], wireless connections, and standardized handling procedures [3, 4]. Still, over the last 12 years, quite a few alternative methods of cardiac triggering have been proposed [7–12], indicating that there is still a need for more reliable and easier-to-use cardiac trigger methods. Some of these methods are based on optical or mechanical measurements. However, the MRI scanner is already a very sophisticated electromagnetic signal generation and sensing device. Therefore, trigger methods like the BioMatrix Beat Sensor (Glossary), which are based on electromagnetic effects are a good match as they can re-use already existing facilities of the MRI machine.

## BioMatrix Beat Sensor – the Pilot Tone

The development of the BioMatrix Beat Sensor started with a power pitch presentation that I attended at the ISMRM 2014 in Milan, Italy. It was about respiratory noise navigation and was given by Anna Andreychenko [13]. It is well known that respiration modulates the electric load of the MR receiver coils. If the receivers are matched to maintain constant signal under load variations, this results in a modulation of the thermal noise power. Andreychenko demonstrated that respiratory navigation can be derived from measurements of the noise power. I was intrigued by this method and tried it out upon my return from the

conference. I quickly realized that the number of noise data points required to characterize respiration is quite high and difficult to achieve during fast measurements like bSSFP. Luckily, I remembered a presentation that Lars Hanson had given at the IDEA developer conference in Freiburg, Germany, back in 2007 [14]. There, he described a system for “burning” electroencephalography (EEG) information into MR images by modulating the information onto a weak RF carrier close to Larmor frequency. The modulation is extracted by the MR receiver together with the MR data, and the EEG data appears as a stripe artifact in the image, or, if the carrier frequency is placed just outside of the frequency band of the image, i.e., in the oversampling region, the MR data remains uncontaminated by the additional signal. I postulated that any body-motion would modulate a constant coherent RF signal and that the achievable SNR per unit time would be higher than for noise navigation. I discussed the idea with a few experts in the factory in Erlangen who agreed that it should be feasible to use the signal for respiratory navigation. Markus Vester suggested the term “Pilot Tone” (PT) for the new motion-sensing method. In communication engineering, a Pilot Tone refers to a reference signal that is transmitted together with the payload signal to characterize the transmission path.

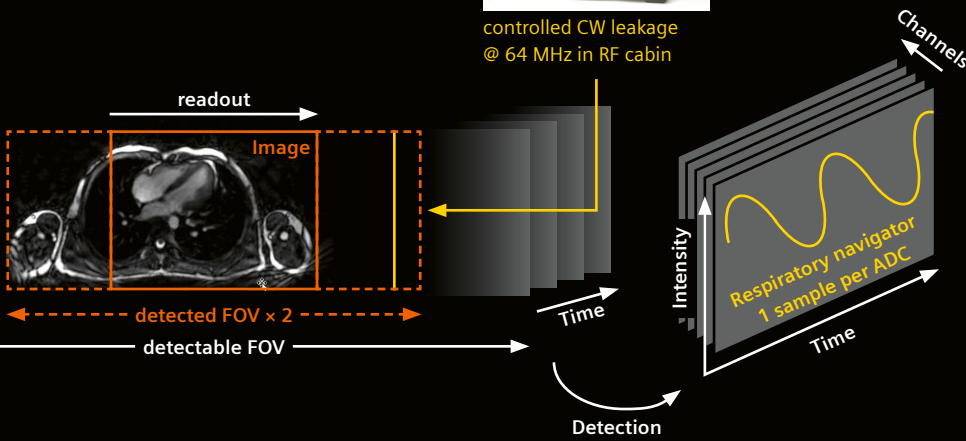
We presented our case and received approval from MR management for an initial cross-departmental investigation. We started the first volunteer tests with an off-the-shelf signal generator in the control room connected through the wave-guide hole to an untuned pickup loop taped to the funnel of a 1.5T MAGNETOM Espree. For the setup, see Figure 1A. The first experiment featured a series of breath-holds alternating between inspiration and expiration. We looked at the data in the command line tool *measdataviewer* directly on the scanner and could immediately see a strong and clean modulation with respiratory position. Over the year, we continued experiments and built an inline-processing module in our image reconstruction environment ICE that would generate live respiratory information and clean up the MR data. I presented the initial results at the ESMRMB 2015 in Edinburgh, UK [15].

**1A Pilot Tone Userland implementation**

- Generate ultra-low-power “leakage” CW signal with fixed frequency just outside MR signal band
- Detect with MR receiver, calc. navigator in recon



controlled CW leakage @ 64 MHz in RF cabin



**1B Beat Sensor – Demonstrator WIP ready\***

Userland (Seq+ICE)

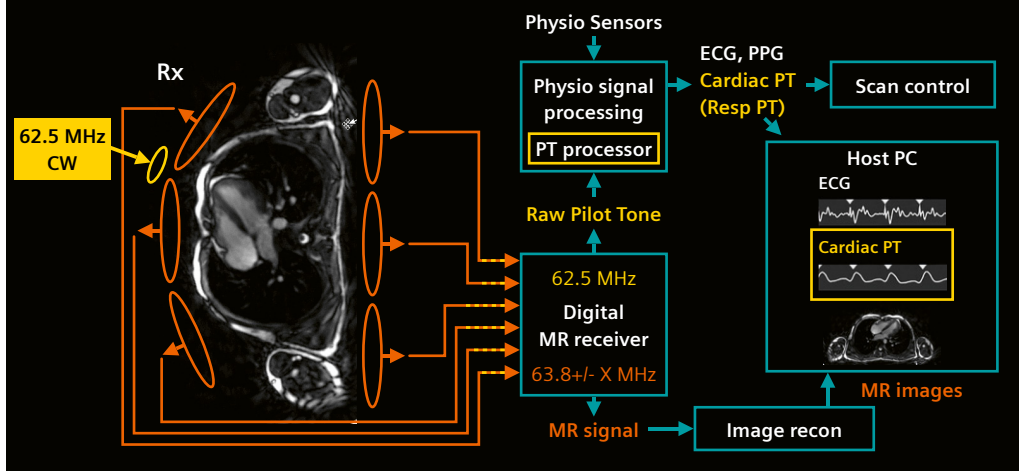


Internal product



\*Active during WIP acquisition only

**1C System integrated Pilot Tone**



- 1** Three integration levels of the Pilot Tone subsystem:
- (1A) Vendor-independent “userland” implementation;
  - (1B) Integrated PT generation and detection for increased signal stability for experimental use by collaboration partners;
  - (1C) Fully integrated PT subsystem to continuously provide physiological information.

Since this abstract is difficult to obtain, it is reprinted with kind permission of ESMRMB in the Appendix.

In summer 2015, we started a collaboration with Fernando Boada's group at CAI2R at NYU. They set up their own version of the experiment on their Biograph mMR and published their first results at the ISMRM 2016 in Singapore [16]. They have continued to develop respiratory PT applications for MR-PET [41, 17] and they even offer their own version of a battery-powered PT generator<sup>1</sup>. Other sites also built their own PT setups for respiratory work and investigated its application to free-breathing MR Fingerprinting (MRF) [18] and prospective slice tracking in free-breathing cardiac MRI [19, 20].

This "userland"<sup>2</sup> approach to PT applications is vendor agnostic, but has three main drawbacks which introduce complications that can reduce their stability:

1. The PT data is acquired only when the sequence plays out an ADC. So, if the measurement is not continuous (e.g., a cine), but contains recovery periods or other pauses (e.g., TSE), no PT data is available in the pauses unless the sequence is modified to fill the gaps with ADCs. However, additional ADCs increase the raw data size significantly. Also, there is no PT data in pauses between measurements.
2. The frequency of the PT must be adapted to the frequency band used by the current protocol: It must fall into the oversampling region that is outside the reconstructed frequency band but inside the oversampled band. The width of the frequency band depends on the protocol (the receiver bandwidth). In addition, the band shifts with in-plane slice shifts in readout direction. The frequency selection becomes easier to fulfill if the oversampling factor (standard is factor 2) is increased, but this also increases the raw data size by the same factor and might not be possible for protocols that utilize high receiver bandwidth.

<sup>1</sup><https://cai2r.net/resources/3t-pilot-tone>

<sup>2</sup>The term "userland" is used here in analogy to the usage in the Linux community to denote functionality programmed only using the vendor-provided programming toolset, for MRI typically limited to sequence and image reconstruction.

### Box 1: A history of electromagnetic monitoring and MR

In 1958, YE Moskalenko, who worked on electroplethysmography in the context of the Russian space program, demonstrated that a radar transmission setup operating at 1 GHz (much higher than the operating frequency of a clinical MRI scanner) could be used to observe respiration and cardiac activity [33].

The first report on an electromagnetic navigation (EMN) signal at low frequency observed in humans was published ten years later by Tarjan and McFee [34]. Their goal was a contactless measurement of blood volume. They adapted a method known in geophysics as an "induction measurement": They projected a magnetic field with a frequency of only 100 kHz into the chest of a volunteer and observed the resulting field with a set of two coaxially aligned receive coils while suppressing direct coil-to-coil coupling. A change in body conductivity would result in a modulation of the received field. They observed a modulation of a few percent due to respiration and a smaller modulation due to cardiac motion of about one percent. They also noted that the signal seems to represent cardiac volume change.

While it has been known in NMR and MRI that an electrically conductive sample loads the coil, and that for optimal efficiency and SNR, the coil must be tuned and matched to the sample, the effect was not exploited for physiologic monitoring until 1988, when Buikman et al. proposed to measure the modulation of the reflection of an RF pulse in the body coil to record respiratory and cardiac information [35].

However, the idea was not picked up widely until quite recently, when new developments enabled robust separation of the different modulation sources: the availability of near-infinite computing power, the availability of massively parallel receive systems on virtually all clinical MRI scanners, and the availability of highly parallel transmit systems on high-field MRI scanners. Where highly parallel transmit systems are available, the reflection of all their elements can be monitored during RF pulses [36]. When measuring the coupling between all elements, high-quality rich data is obtained that allows separation of multiple motion contributions [37], but only while transmitting RF pulses.

The noise navigator [13] and the Pilot Tone [15] require multi-channel capability on the receive side only. Therefore, they can be implemented on all modern MRI systems and while the MR receiver operates, navigation data can be obtained.

The most recent development in EMN for MRI is the "Beat Pilot Tone" [38]: The method removes the requirement that the PT frequency be near the Larmor frequency by generating the signal close to the Larmor frequency only in the MR receive chain as an intermodulation product of two electromagnetic signals with higher frequencies, where these two frequencies are separated by approximately the Larmor frequency. This approach allows low-cost integration of EMN at radar frequencies into the MR scanner by using the MR receive system as a continuous wave (CW) radar detector.

3. The PT appears as a peak in the Fourier-transformed ADC. The peak position is a function of the PT frequency and the scanner's detection frequency; thus, it depends on the slice geometry, i.e., on the off-center shift in readout direction. This correction is deterministic and can be calculated exactly from protocol parameters and raw data headers. However, the eddy current compensation contains a  $B_0$  correction, which is realized by varying the receive frequency. The correction thus generates additional, small transient shifts in the PT peak position when gradients are applied. These will become noticeable when the eddy current steady state changes markedly, e.g., when toggling between slice orientations. To avoid this problem, the PT peak position must be redetermined with every substantial change in the gradients, e.g., when switching slice orientations or when playing out large phase encoding jumps. In addition, the PT frequency from a generator that is not synchronized to the MR scanner will drift with time, and the detection frequency must be adapted if the drift is too large.

To eliminate these problems, we integrated the Pilot Tone generation into the MAGNETOM architecture as a prototype. The solution was to generate the PT at a fixed frequency and detect it directly after digitization before the "userland" signal with all corrections is derived. This made it necessary to implement the PT detection in firmware. To save field-programmable-gate-array (FPGA) resources we picked as frequency Nyquist/2 in the digitized frequency band, where the detection algorithm would not require multiplications but only additions. This led to the currently used PT frequency 62.5MHz @ 1.5T. To avoid drifts of the PT signal in frequency and phase, and thus enable complex valued processing, the PT signal is generated by the scanner itself using a "spare" frequency generator.


In parallel, we continued to explore the possibilities of the new method in a master's thesis in the cardiac predevelopment team, together with the local university [21]. Our department head, Lars Lauer, challenged us to extend the scope of the thesis beyond respiratory navigation and aim for cardiac triggering. At that time, we had no indication that this would be feasible, as we had never seen a cardiac signal with our remote generator set-up. If we had studied the scientific literature better (Box 1), we might have been more hopeful, but this only happened in the following master's thesis [22]. Halfway into Lea Schröder's thesis, Jan Bollenbeck built a battery-powered PT generator that could withstand the RF pulses of the MR experiment. The aim was to place the generator closer to the patient's chest to reduce sensitivity to non-respiratory motion, e.g., head or foot movement. We first attached the generator to the scanner's inner bore cover over the chest and didn't see anything new. However, when we placed it directly on the chest of the volunteer, we could see the cardiac signature directly in the raw data. These results were presented at ISMRM 2016 in Singapore [23] (Fig. 2). Internally, they also raised a lot of interest because of the potential for developing the new method into an alternative to ECG.

These initial results were seen promising enough to start the product integration. The first goal was to enable collaboration sites to start their own PT experiments with an integrated, stable, and easy-to-use system that featured local PT generation. I was able to convince Peter Gall, the product manager of the new system platform that was in the planning stage at that time, to include the necessary hardware (HW) and software (SW) in the system project.

Early in 2016, we started a second master's thesis on PT, this time with the Graz University of Technology in Austria, to further develop the cardiac application of the PT [22]. Using a second generation of the battery-powered PT generator, Mario Bacher ran a volunteer cohort study and developed signal processing SW for retrospective and

### A novel method for contact-free cardiac synchronization using the PT navigator

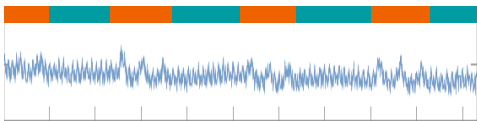
**Local PT navigator**



Battery-driven autonomous RF source placed on chest

... „sees“ respiratory motion + something else

PT navigator example for instructed breathing:  
Interleave 3 deep breaths and breath hold



Question: *Is this cardiac motion?*  
Task: *Separate signal contributions*

**2** The key slide from [23] demonstrates that cardiac motion can be detected using a locally generated Pilot Tone.

## Box 2: PT physics: The transformer model

The PT signal, like all electromagnetic applications, is a result of the Maxwell equations ;-). To understand real-world scenarios, especially complex ones like humans, numerical simulations are required. However, insight into the basic principles can be obtained with idealized, simple models: the main mechanism of the PT modulation by cardiac motion can be understood by comparing the setup to an electric transformer. The pre-condition is that the PT-generating loop is much smaller than the electromagnetic wavelength and thus mainly generates a magnetic field and only a negligibly small electric field. Thus, the PT is a purely magnetic signal which exists only as a near field; contrary to radar, which is a self-propagating electromagnetic wave, used in its far field.

A transformer consists of two coils: The primary driving coil (yellow, left side) produces an oscillating magnetic field. The resulting flux through the second coil (yellow, right side) induces a voltage. The flux between the coils is maximized by connecting the two coils with a magnetic, but non-conductive medium. Luckily for us MRI people, the human body is only weakly magnetic and it conducts electricity quite well. Therefore, the body can be seen as a magnetically transparent but lossy transformer core. The primary magnetic field (blue) will generate eddy currents in the tissues (orange), depending on the local conductivity. These eddy currents in turn generate secondary magnetic fields (orange). The secondary field opposes the primary field at the point of origin. In this example we observe the superposition of both fields in the receive coil on the right where both field lines point in different directions. Thus, depending on the geometry

the secondary field can cause decreasing or increasing amplitude and/or phase shifts.

The body consists of tissues that have different electrical conductivity and change their position during motion. When conductive tissues move, the eddy current distribution changes and the net secondary field with it. The resulting modulation of the received fields is observed as the Pilot Tone signal. Even though the heart is relatively small, its motion is clearly observable because during cardiac contraction, highly conductive heart muscle and blood are replaced by weakly conductive fat and lung tissue.

As mentioned earlier, the Pilot Tone frequency needs to be chosen fairly close to the Larmor frequency: 62.5 MHz for 1.5T systems. Again, we were lucky: These frequencies hit a sweet spot where on the one hand wavelengths are long enough (2.4 m to 4.8 m in a vacuum, a few decimeters in tissues) so that the signal penetrates the body quite freely and can therefore be described by this simple model, and where, on the other hand the frequency is high enough that the induced eddy currents and consequently the secondary magnetic fields are strong enough to be easily detected.

All these factors together enable reliable detection of cardiac, respiratory, and even head motion at standard clinical field strengths. However, as Anand et al. [38] have shown, exciting opportunities also await at very high frequencies! When venturing into the GHz range, the transformer model is no longer applicable, as we are entering the domain of radar. Here, higher order interactions start to dominate the signal and limit its penetration into the body, which is expected to optimize the detection of rigid motion.

### How does it work?

#### Transformer model

Assumption<sup>1</sup>:  $B \gg E$

A transformer consists of

- Primary **driving coil** →  $\vec{B}$
- Secondary **receiving coil**
- A (highly) magnetic but non-conductive medium

If medium is conductive → eddy currents  $i_e$

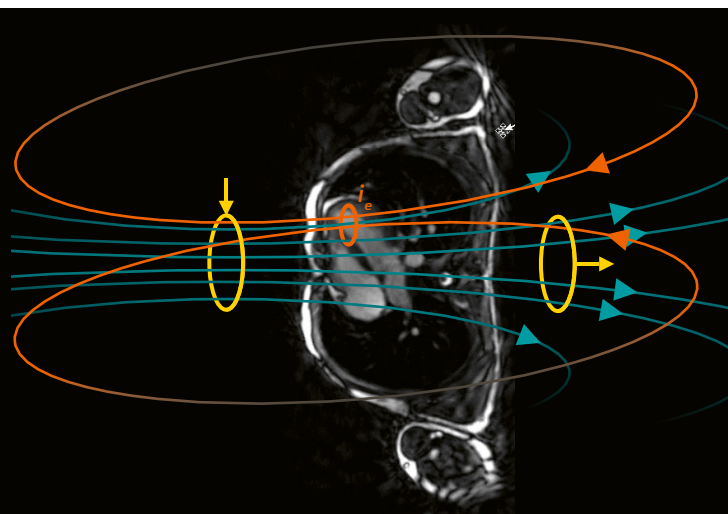
→ Opposing  $\vec{B}_e$

Conductivity  $\sigma$ : Blood > Muscle > Lung > Fat

Motion changes EC distribution

→ received signal changes in amplitude **and** phase

<sup>1</sup>Ensured by small generator loop



finally also for prospective cardiac triggering (Box 3). He obtained his first prospectively triggered cardiac cine still with a battery-powered generator and a pure “userland” implementation. He went on to work on a PhD thesis on the topic at Lausanne University Hospital (CHUV) in Switzerland. In his work, he also unearthed the early history of electromagnetic navigation and its use in MRI (Box 1), and the physics of signal generation (Box 2).

For the first version of the integrated PT-generation and detection system, (Fig. 1B) the PT was generated inside the new BioMatrix 12 coil. The PT was still detected only during ADCs and evaluated in ICE, but it was already much more stable. In this context, the term “Beat Sensor” was introduced to describe the cardiac application of the Pilot Tone (Glossary). This system was subsequently used at the Royal Brompton Hospital (RBH) in London, UK, to acquire the first cardiac PT patient data [24], and at CHUV and Ohio State University (OSU), Columbus, OH, USA, to develop and evaluate respiratory and retrospective cardiac applications (see e.g., Falcão [25] and Chen [26]).

Developing a robust and complete cardiac PT application takes time, and we went through several releases that gradually moved us closer to the current product. First, we implemented a general PT framework (see Fig. 1C) with the following main features:

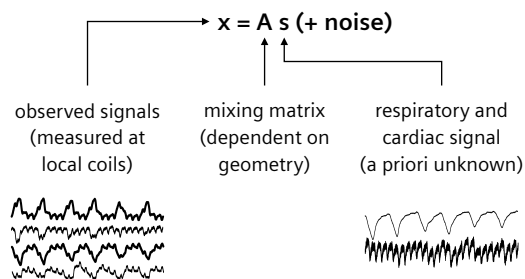
- An acquisition-independent and continuously acquired PT data stream, with samples being flagged as invalid during RF pulses
- Pairing each PT sample with information about the preceding RF pulse to enable spike subtraction
- A PT-processing framework that can be exchanged and reconfigured for individual measurements to enable rapid prototyping and work-in-progress packages.

The first point amounts to quite a substantial paradigm shift: Before, the MR receiver had to be configured and active only during measurements. Now, the receiver had to be always configured and active to receive PT data during and between measurements. To achieve this, a dedicated team spent much of the development time

### Box 3: Signal processing basics

A key advantage of the PT method is the use of all connected coil elements for receiving the PT signal. Each receive coil “sees” a different combination of cardiac, respiratory and other motion, depending on its location. Since we are interested in observing cardiac and respiratory motion by themselves, these components must be separated first.

This can be accomplished by a class of algorithms known as “Blind Source Separation”. The problem statement is simple: Given a multitude  $N$  of sensors measuring signals  $x$  which are a combination  $A$  of  $M < N$  independent signals, separate the underlying signals  $s$ .



In other words: the equation needs to be solved for  $A$ , the so-called mixing matrix. Initially we used FastICA (Independent Component Analysis, [39]), in which the non-Gaussianity is optimized as a proxy for statistical independence [22]. While this algorithm worked well for Cine sequences, which are not affected by RF inter-

ference, a new PCA-based algorithm was developed in collaboration with the Shenzhen team, based on [27], in which the RF contamination of the PT signals is minimized by mixing vectors that are orthogonal to the main principal components of the eigenspace of interference signals.

To enable robust cardiac triggering, the cardiac motion signal needs to be de-noised in real-time. Classical digital filters, however, would introduce an unacceptably large time delay for the cardiac application (e.g., > 600 ms for an 80 dB stopband FIR lowpass at 6 Hz).

In [12, 22], we proposed to use a Kalman filter to denoise the signal. A Kalman filter, named after Rudolf E. Kalman, is an algorithm that uses a series of measurements, including noise and other inaccuracies, and attempts to fit these measurements in a least-squares sense to a predefined signal model. In our case, the signal is modeled using a constant velocity motion model similar to Spincemaille et al. [40], in which the signal evolution is modeled as the 1D motion of a particle with constant velocity. This approach has two additional advantages: First, it does not only provide a de-noised signal, but also an estimation of the signals first derivative, i.e., velocity, which is subsequently used for trigger detection (see Box 4). And second, it can interpolate over short gaps in the data that occur while sending RF pulses.



on writing and debugging the HW–SW interconnectivity. The flexible processing framework drastically sped up the subsequent development by enabling us to test new code on previous stable baselines and SW versions. To minimize porting efforts, we kept interfaces constant and minimized dependencies on other modules.

In 2021, we released the first clinical application: Beat Sensor triggering for Cines only. The limitation to Cine was necessary because we had not yet implemented a general RF-artifact compensation, and for Cine a simple signal average is sufficient. Most of the development time was spent on optimizing signal processing and workflow. For about one year we had regular volunteer shifts to test stability and usability of the latest developments. A lot of time was spent on the special coil handling required for Beat Sensor-triggered measurements: The PT signal is acquired using the coil elements currently selected for imaging, and signal-processing training is valid for one

coil select only. Therefore, Beat Sensor-triggered protocols must automatically select the coils used in training.

While the team in Erlangen, Germany, had been working towards productization of the cardiac application of PT, the Siemens Healthineers development team in Shenzhen, China, had been developing the respiratory application of PT. After an initial ramp-up phase, the team in Shenzhen independently developed and released a first product for respiratory triggering on MAGNETOM Amira – A BioMatrix System. After this release, the two groups intensified their collaboration: Yan Tu Huang in Shenzhen integrated his respiratory processing into the PT framework, and several algorithms were exchanged between the sites and applications. The cardiac application benefited most by rebasing on the robust method for RF-artifact suppression [27] that Shenzhen had developed for the respiratory application. We jointly extended this algorithm for the cardiac application. Based on this algorithm we were able to stabilize

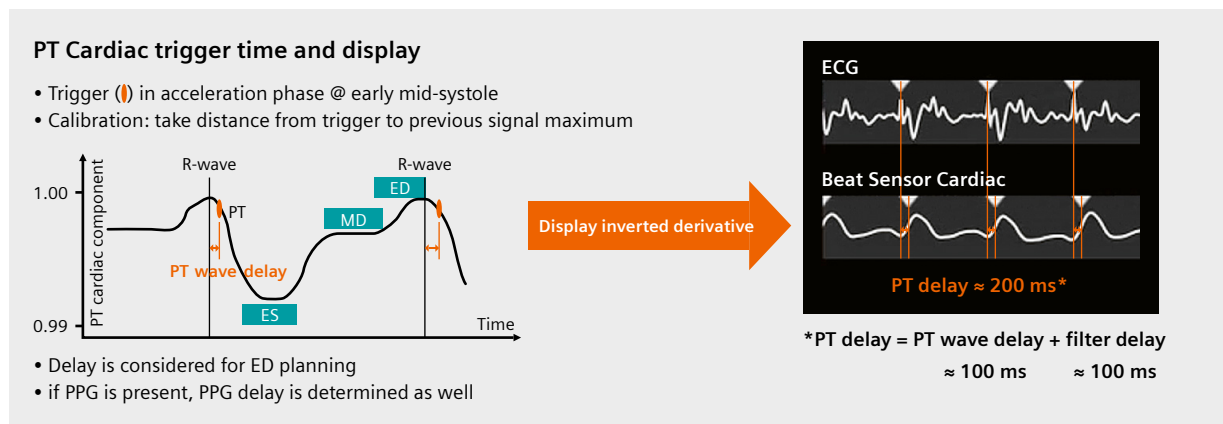
### Box 4: PT signal characteristics and trigger time point

The cardiac component of the Pilot Tone closely follows the cardiac volume curve without a delay: At the time of the R-wave, in end-diastole, when the heart is fully expanded, it assumes its maximum value. During end-systole, when the heart is fully contracted, it assumes its minimum value, and during the static mid-diastolic period it plateaus at an intermediate value. The Beat Sensor signal, shown on the right, is the inverted derivative of this cardiac component. It has the advantage that it features a rather narrow positive peak at the beginning of the cardiac cycle and provides additional suppression of respiratory contamination.

The Beat Sensor triggers when the systolic contraction starts (specifically, when the derivative reaches 40% of its maximum value). This trigger point was chosen for its stability and occurs approximately 100 ms after the R-wave. This wave delay is learned during the

Beat Sensor training phase using only the waveform of the cardiac component as the distance between trigger point and the preceding maximum. Low-pass filters in the signal processing add roughly 100 ms to this, resulting in a total delay of about 200 ms. If a finger pulse sensor (PPG) is present as well, its delay with respect to the R-wave is also determined.

One crucial functionality for cardiac exams is “Capture Cycle”: based on the current heart rate, an algorithm places the acquisition in the end-diastolic cardiac phase where motion is minimal. For the ECG, that triggers on the R-wave, this placing is achieved by maximizing the trigger delay. For Beat Sensor and PPG triggering, the algorithm has been modified to account for the trigger delays learned during Beat Sensor training. Thus, after the training “Capture RR” can be used for all three trigger sources.

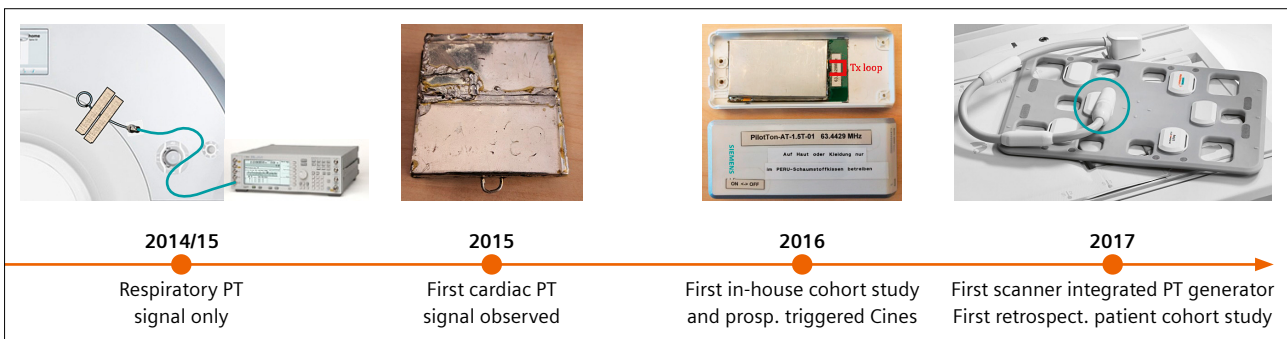


cardiac triggering for all cardiac protocols. The signal characteristics of the current implementation are described in Box 4.

To collect customer feedback early, we developed a work-in-progress package (WIP) of Beat Sensor triggering for the whole cardiac exam. The WIP was initially installed at four different sites (Jan Yperman Hospital, Ieper, Belgium; Royal Brompton Hospital, London, UK; Ohio State University, Columbus, OH, USA; and Northwestern University (NWU), Evanston, IL, USA). While the feedback on clinical performance was positive [28–30], the calibration workflow proved to be too complex and error-prone. Thus, we reworked and hardened the calibration workflow just before the development deadline and could release Beat Sensor triggering for the whole cardiac exam for the current SW version *syngo* MR XA51A for the 1.5T BioMatrix Systems MAGNETOM Sola, MAGNETOM Altea, and MAGNETOM Sola Fit. The next SW version for 3T systems will bring this functionality to MAGNETOM Vida and MAGNETOM Lumina.

The development of PT-based methods is ongoing. The clinical use of the first complete release of the Beat Sensor will certainly reveal limitations, but it will also show distinct advantages when compared with the ECG. We are committed to removing limitations and turning the advantages into clinically useful features. So please try out the Beat Sensor in your clinical routine and let us know about your experiences.

Looking beyond cardiac to general motion management, the use of Pilot Tone is still in its infancy and we have only scratched the surface of many aspects and applications. Two examples are two-dimensional characterization of respiration [31] and characterization of head motion [32]. The scanner-integrated PT with its continuous data stream and its flexible processing framework is a powerful tool to enable these upcoming investigations and developments.



**3** Four generations of Pilot Tone generators: From initial respiratory experiments to a scanner-integrated solution.

## Acknowledgment

A big Thank you to all contributors. Especially:

- **Generator hardware:** Jan Bollenbeck
- **Physio-SW:** Michael Schwertfeger
- **PT processing:** Mario Bacher, Yan Tu Huang
- **Sequences, exam prototypes and data analysis:** Carmel Hayes, Randall Kroeker
- **Volunteer data:** Manuela Rick
- **Early patient data:** Peter Gatehouse (RBH)





## Appendix: Reprint of Ref. 15

### PT-Nav: a novel respiratory navigation method for continuous acquisitions based on modulation of a pilot tone in the MR-receiver

P. Speier<sup>1</sup>, M. Fenchel<sup>1</sup>, R. Rehner<sup>2</sup>

<sup>1</sup>MR PI, Siemens Healthcare, Erlangen, Germany

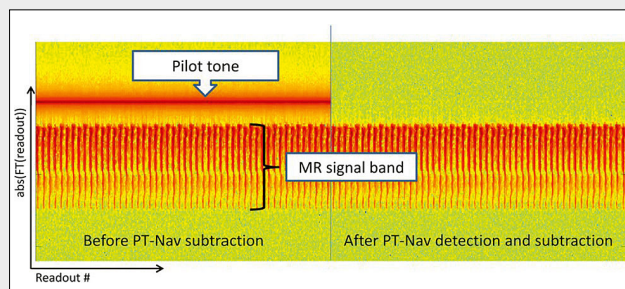
<sup>2</sup>MR R&D, Siemens Healthcare, Erlangen, Germany

#### Purpose/Introduction

High-resolution 3D abdominal or cardiac MR imaging requires synchronization to respiratory motion. The motion can be tracked by dedicated respiratory sensors, MR navigators, selfgating algorithms, or, as recently proposed, by analyzing the received noise [1] or measuring load changes of the transmit coil [2]. These methods increase setup complexity, require interruption of steady state, special k-space trajectories and high SNR per scan, long averaging times, or additional scanner-integrated hardware. We hypothesize that a coherent signal from an independent source, received in parallel with the MR signal, can be used as a high-quality navigator that does not suffer from the drawbacks of the previous methods.

#### Subjects and methods

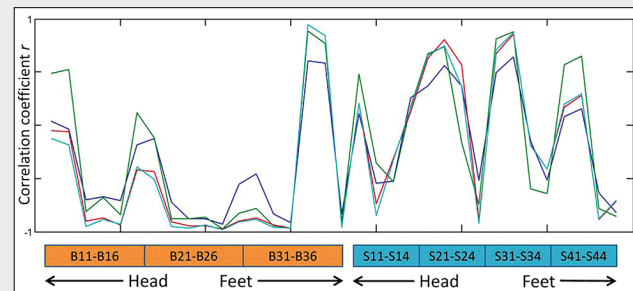
Continuous series of sagittal bSSFP or GRE images (500...1000 images, 3...5 images/s, (1.5...2 mm)<sup>2</sup> × 10 mm) of the right liver dome were acquired with anterior and posterior multi-channel receive coils. Meanwhile, a continuous-wave (CW) RF signal, generated by a commercial signal generator, was transmitted as a pilot tone into the magnet bore by a non-resonant pickup coil, placed on the table or outer magnet cover, with a frequency outside of the frequency band of the MR signal and the FOV, but inside the received frequency band, and an amplitude



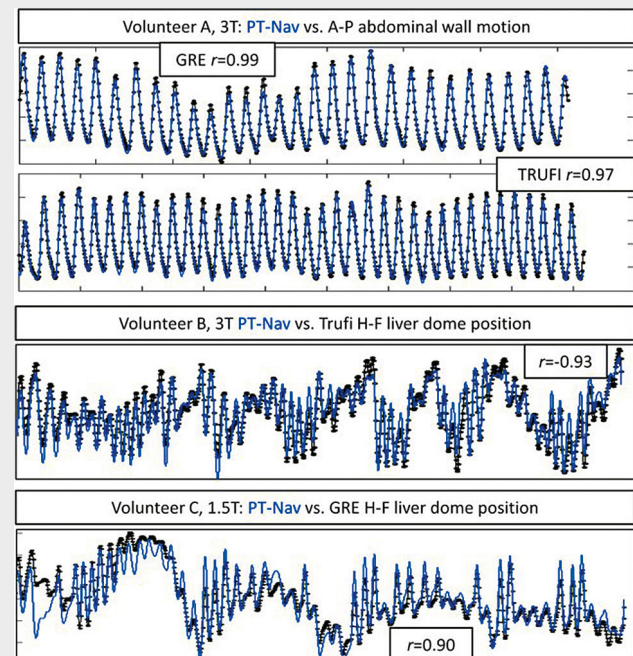
**1** Exemplary Fourier-transformed raw data (one channel), demonstrating successful detection and elimination of a pilot tone placed next to the MR-signal band

adjusted to be detectable in the linear operating regime of the receiver.

Image reconstruction was modified in-house for pilot tone detection: During a learning phase, the pilot tone frequency  $f$  in the received data was determined. Then, for each k-space line and channel, the pilot tone was fit to the model  $A \times \exp(i2\pi ft)$ , the complex amplitude  $A$  logged as PT-navigator, and the model subtracted from the data (Fig. 1). Offline processing was performed in MATLAB (Math-Works). The positions of the liver dome (HF direction) or abdominal wall (AP direction) were tracked in the images, and their correlations to low-pass (1 Hz-Hann-window)-filtered PT-navigators were calculated for all channels.



**2** Exemplary correlation coefficients of anterior (orange) and posterior (blue) coil elements for 4 measurements of volunteer B



**3** Exemplary correlations between best PT-Nav channel and image-based navigators (curves shown after mean subtraction and amplitude normalization)

## Results

Measurements of 5 volunteers were performed on three systems (MAGNETOM Espree, Aera, and Skyra, Siemens Healthcare). Correlation factors varied in sign and amplitude between channels (Fig. 2). For all volunteers, receive channels with high correlation coefficients  $r \geq 0.9$  were found while volunteers laid still, allowing precise characterization of even quite irregular respiratory patterns (Fig. 3). Correlations could be slightly improved by combining channels.

## Glossary

### Electromagnetic navigation (EMN)

Characterizing motion by its influence on low-frequency electro-magnetic fields inside the body, used here with a focus on scanner-integrated methods

### Pilot Tone (PT)

MR-integrated EMN, the only hardware part is the field generator

### BioMatrix Beat Sensor

1. Cardiac application of Pilot Tone
2. Hardware integrated Pilot Tone generator (BioMatrix 12, BioMatrix 18 coils)

### BioMatrix Respiratory Sensor

1. Respiratory Sensor hardware in BioMatrix Spine coils on 1.5T MAGNETOM Sola, and MAGNETOM Altea, as well as on 3T MAGNETOM Vida, and MAGNETOM Lumina (this is EMN, but not PT based)
2. Respiratory PT application on 1.5T MAGNETOM Amira - A BioMatrix System (uses a PT generator in the BioMatrix 13 coil)

## Discussion/Conclusion

Our early results suggest that the proposed PT-Nav method can provide, with minimal hardware requirements, respiratory information for continuous sequences that is comparable to that of "gold-standard" MR navigators.

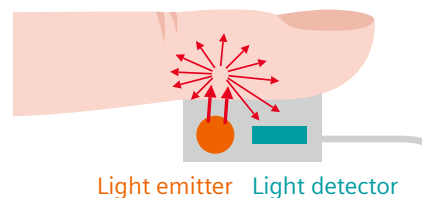
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### Why is "Beat Sensor" an appropriate term for the Pilot Tone generator?

Like the finger pulse sensor, the Beat Sensor is an active sensing method: It emits a signal into the body and receives the motion-induced modulation of this signal. The Beat Sensor hardware, that is built into the BioMatrix 12 and BioMatrix 18 coils, fulfils the function of a signal generator but does not receive the signal. It can be likened to the light emitter in the finger pulse sensor. The role of the light receiver is implemented by modifying the SW of the already existing MR receive system.

HW	Pulse Sensor	Beat Sensor
Generator	Light emitter	PT generator: dedicated hardware in BioMatrix 12 and BioMatrix 18 coils
Receiver	Light detector	PT receiver (reuse MR receive system)



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## Contact

Peter Speier, Ph.D.  
Cardiovascular Application Development  
SHS DI MR RCT CLS CARD  
Allee am Roethelheimpark 2  
91052 Erlangen  
Germany  
peter.speier@siemens-healthineers.com

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Watch as BioMatrix Beat Sensor developers Peter Speier, Mario Bacher, Carmel Hayes, and Manuela Rick explain the various stages of the development.



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