Fiber-Optic Cardiorespiratory Monitoring and Triggering in Magnetic Resonance Imaging

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***Abstract*—During the past decades, fiber-optic technology has become a very popular tool for vital signs monitoring. Thanks to its advantageous properties, such as noninvasiveness, biocompatibility, and resistance to electromagnetic interferences, this methodology started to be explored under the conditions of a magnetic resonance (MR) environment. This review article presents the motivation and possibilities of using fiber-optic sensors (FOSs) in MR environment and summarizes the studies dealing with experimental validation of their compatibility with MR. Several aspects of the presented issue are highlighted and discussed, such as suitability of the fiber-optic approach for MR triggering, precision of vital sign detection, development of sensor designs, and its application to patient's body. From the literature review, it can be concluded that FOSs have promising future in the field of cardiorespiratory monitoring in MR environment. This is mainly due to their advantages originating from sensing mechanical signals instead of electrical ones, which makes them resistant to MR interference and extrasystoles. Moreover, these sensors are easy to use, reusable, and suitable for combined monitoring. However, there are several shortcomings that should be solved in future research before introducing them to clinical practice, namely, signal's delay or optimal placement of sensors.**

***Index Terms*—Cardiorespiratory monitoring, fiber-optic sensors (FOSs), magnetic resonance imaging (MRI), MRI triggering, vital sign monitoring.**

I. Introduction

In recent years, magnetic resonance imaging (MRI) has progressed significantly, as it now provides invaluable tissue composition, functional, and metabolic structure information. Simultaneously, advances in medical computing have enabled the advent of many advanced quantitative techniques for enhanced disease diagnosis, prevention strategies, and administration of therapy. The development of higher magnetic field MRI machines, nowadays reaching to 7 T, and their growing deployment in clinics have brought about new challenges in MRI imaging. One such important and significant challenge is monitoring and compensating for movements resulting from patient's cardiorespiratory activity during imaging [1]. One such compensation method is timed-triggered imaging which is paramount for obtaining quality diagnostic information [2]. Such adjustments are even more essential for a relatively large percentage of patients who are noncooperative such as neonatal, pediatric, and mentally or critically ill patients [3]. However, the most commonly applied adjustment methods are adversely affected by high electromagnetic interference, acoustic noise, and vibration. These effects become even more prominent with the increase in the MRI's magnetic field strength [4], [5].

Fiber-optic sensors (FOSs) are advantageous to use in medical field, thanks to their noninvasive nature, biocompatibility, high sensitivity, small dimensions, and reduction of wires used. Since they are composed of dielectrics such as glass or plastics, the electrical isolation of the patient or elimination of voltage induction is not needed [6]–[9]. Moreover, their construction is characterized by high resistance to electromagnetic interference, which is suitable for making measurements in magnetic resonance (MR) environment [10], [11]. The FOSs use the changes in the light properties of the applied light (such as wavelength or frequency) due to cardiorespiratory or other physiological activities to detect the presence of such activities. Indeed, with this method respiration, heart rate (HR), blood composition, and body temperature can be monitored.

Previously, the role of FOSs in monitoring of cardiorespiratory activity and other vital signs and, particularly, their suitability for patient monitoring, including their integration into smart textile, have been reported [12]–[18]. This review article focuses on recent advances in monitoring of patient's cardiorespiratory activity during MRI with the option of MRI synchronization. This problem has been already outlined in our previous article [19] that provides a comprehensive summary of methods suitable for MRI gating and their comparison. In contrast, the focus of this article is more in-depth and only includes the overview of the sensors based on fiber optics that were successfully tested and validated in MR environment. It presents the application of various types of FOSs and discusses the finding of their latest experimental evaluations.

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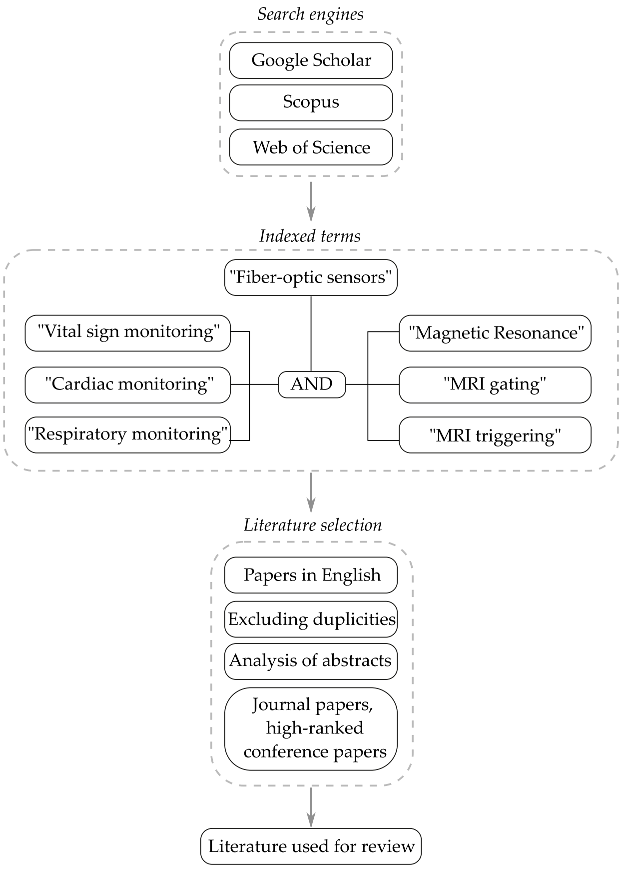

Fig. 1. Selection of literature for the review and indexed terms.

This literature review was made by exploring full articles, which include original journal articles, conference proceedings, review articles, and book chapters in English language. As the review focuses on a quite narrow topic, the indexed terms and their combinations had to be selected appropriately to achieve a sufficient spectrum of results (see Fig. 1). For the selection of suitable literature sources for this review, Scopus and Web of Science databases were preferred as they provide a wide range of peer-reviewed sources from various technical fields, including biomedical engineering. From the articles collected in this way, duplicities were excluded and the remaining articles were investigated. The publications dealing with fiber-optic technology for cardiorespiratory monitoring in MRI were selected for further analysis.

II. Vital Sign Monitoring in MRI

In the context of MRI clinical applications, among the involuntary human activities that can be noninvasively measured, cardiorespiratory activity is the most important. The importance of monitoring this activity is two folds. First, it can reveal the state of the patient's wellbeing, and second, it provides an opportunity to improve the quality of MRI scans that can otherwise be distorted due to involuntary motions associated with heartbeat and respiration. In addition, more useful, and effective information about patient's cardiorespiratory health may be extracted from the measurement of these activities. Examples of such additional information include detection of significant peaks, respiratory rate (RR), HR, and HR variability (HRV).

The vital sign monitoring during MRI can be divided into two main categories according to their intended purposes.

1) *Monitoring of the Patient's State*—There are several patient conditions for which monitoring vital signs is important. Two prime instances of such conditions are seriously ill patients or those in critical condition and patients who suffer from claustrophobia or panic attacks. While a patient who is seriously ill or is in critical condition normally has entered such a state prior to being scanned by MRI, claustrophobia or panic attacks can occur instantly during scanning. Typically, such reactions are underdiagnosed or caused by the patient having to be placed in a narrow cylindrical tube for dozens of minutes and not being allowed to move body parts (especially head). Although usually administering anxiolytics and sedatives to the patient are effective in addressing these stressful conditions, in certain cases it may not be advisable or safe [20], [21]. Moreover, vital sign monitoring is particularly important since these disorders are often undiagnosed and can develop instantly during MRI examination. Monitoring the cardiorespiratory activity allows the clinician acquiring the MRI scan to objectively evaluate patient's symptoms, such as tachycardia or hyper ventilation (see upper part of Fig. 2) and decide whether to terminate the MRI scan to halt the progression of a panic attack [22].
2) *MRI Triggering*—Another important reason for cardiorespiratory monitoring during MRI scan is synchronization or gating/triggering, which enables one to acquire a high-quality image by eliminating any undesirable artifact caused by breathing and heart activity. The principle for such an enhancement is founded in the detection of significant peaks in the measured signal. Using this information, one can select the MRI signal acquisition time to occur at the desired temporal location during cardiac and respiratory cycles, most often at the end of diastole and expiration (see lower part of Fig. 2). Such time-triggered images are characterized by higher contrast of the tissues of interest, contrast of contours, signal-to-noise ratio (SNR), and both temporal and spatial resolutions. Furthermore, this form of triggering provides a scan of the heart in its specific phase of cycle that could be an important diagnostic parameter [19], [23], [24].

A. Conventional Methods

In current clinical practice, various methods of measuring cardiorespiratory activity are used. As necessitated by the strong electromagnetic field of MRI, all the electrodes or other sensors need to be free from ferromagnetic parts to avoid any high-voltage induction that can hurt or even burn the patient [25], [26]. A brief overview description of these methods and their advantages and disadvantages are presented below.

1) *Electrocardiography (ECG)* is considered to be the gold standard in measuring heart activity in many fields. It enables a quick recording by placing multiple

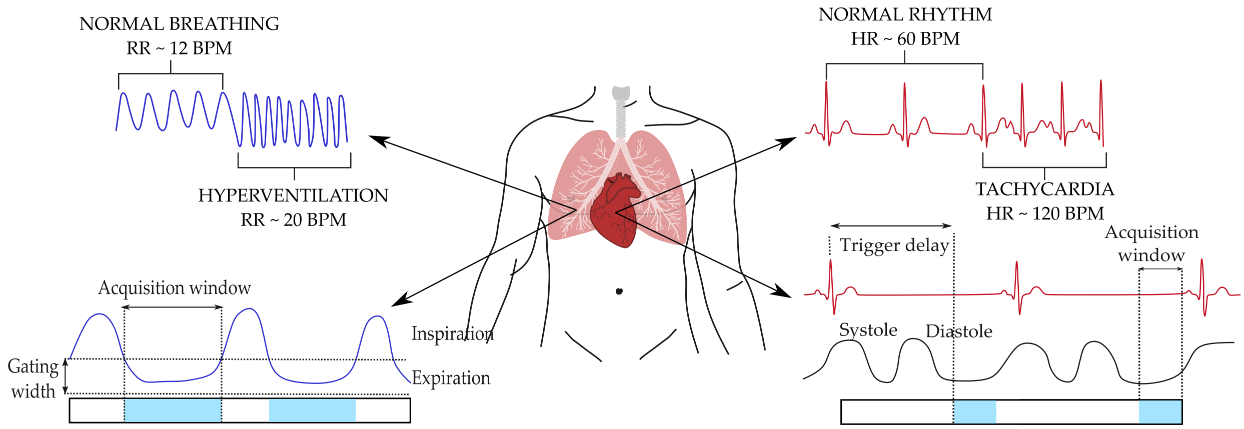

Fig. 2. Cardiorespiratory parameters monitored in MRI (upper) and principle of MRI triggering (lower).

(up to 12) electrodes on the patient's chest. ECG provides detailed information about the individual's HR and cardiac cycle. For measuring during MRI, three or four leads are sufficient to capture the electrical activity of the heart and determine HR. However, the preparation and placement of electrodes is rather involved when compared with nonelectrical measurements. For instance, it may require removal of the body hair at the electrode placement locations and the skin surface may need to be scrubbed off with a mild abrasive soap. Insufficiently prepared skin causes higher skin impedance or artifacts due to poor electrode contact with the skin [24]. The most significant limitation of ECG usability in MR environment is spatial and temporal distortions caused by strong electromagnetic field (i.e., magnetohydrodynamic artifact). Particularly, when using high MR field strengths (beyond 3 T), the signal is not usable for intended purposes [27]–[29]. According to several studies, misdetection of the fiducial points can reach up to 30% of all cases [30], [31].

2) *Vectorcardiography (VCG)* is based on the same principle as ECG but represents the cardiac activity in 2-D or 3-D loop reconstruction, where each loop represents specific signal feature. In some clinics, VCG is the standard method used for MRI applications since the signal is more resistant to magnetohydrodynamic artifact than ECG due to the opposite orientation of the electromagnetic field and VCG fiducial points [32], [33]. However, this resistance decreases with the increasing magnetic field strength, and in 7 T, VCG signal is affected like ECG [34].
3) *Pulse Oximetry (POX)* is widely used in clinical practice for monitoring the patient's state. For MRI triggering, it is a compensatory method used only when ECG cannot be measured for some reason. Pulse wave recording is based on assessing the blood absorption of light corresponding to the changes in blood volume, which makes the method resistant to MR interference [35]. An advantage of POX method is its ease of application, as the sensor is placed on the patient's finger. However, the probe is very susceptible to movement. Moreover, the delay between time of cardiac activity (i.e., mechanical ejection of blood from left ventricle) and sensing of the resulting pulse is several 100 ms [30] which is undesirable for MRI triggering.
4) *Respiratory Belts* make use of piezoelectric sensors to measure respiratory curve from the patient's abdomen or chest, depending on the prominent site of breathing-induced movement in the upper torso. The benefits of the method are not getting affected by the magnetic field strength [36] and providing continuous signal. However, fastening the belt to the patient body can be problematic because the belt must be wrapped around the patient's upper torso tightly to provide accurate detection of breathing. Furthermore, the belts are susceptible to patient's movement [37].
5) *Respiratory Navigators* are used for triggering in MRI of abdomen and are based on transmitting the additional radio frequency pulses whose echo's detection enables monitoring of the movement of diaphragm. A respiratory navigation system helps increase patient comfort. However, there is the possibility of the navigator pulse getting affected by the MR signal which can decrease the image quality. Also, this method is not entirely independent of the magnetic field strength (3 T and beyond) [36]. In cardiac MRI, navigator triggering is not a suitable method, as it allows acquisition during a short sample window within each heart cycle (cardiac MR data need to be acquired throughout the whole heart cycle) [38].

B. Alternative Approaches

In recent years, some alternative approaches to sensing of cardiorespiratory activity have been proposed that do not use fiber optics. A brief review of these methods is presented here to provide a more inclusive comparison of the methods with FOSs that are described in this article. One can observe that the main benefits of measuring using FOSs lie in its high sensitivity, indifference to magnetic field environment, and its ability to sense both the respiratory and cardiac signals. In contrast, most standardly used triggering systems are known to be more or less sensitive to the increasing MR field strength, which makes its use limited in ultrahigh fields [19]. These limitations are summarized in Table I.

TABLE I
SUMMARY OF GENERAL SPECIFICATIONS OF METHODS USED FOR MR GATING

Method	Cardiac	Respiratory	MR strength sensitive
ECG, VCG	Yes	No	Yes
POX	Yes	No	No
Respiratory belts	No	Yes	No
Navigators	No	Yes	Yes
PCG, SCG	Yes	No	No
DUS	Yes	No	No
Self-gating	Yes	Yes	No
FOS	Yes	Yes	No

1) *Phonocardiography (PCG)* involves acoustic manifestation of heart activity, known as heart sounds. They are measured by a sensor (usually a microphone) placed on the patient's chest. The use of a single sensor reduces the number of cables and helps in preventing the induction of high currents. The studies have shown high accuracy of PCG triggering in magnetic field strength up to 7 T. However, the accuracy must be secured by appropriate filtration since the PCG method is susceptible to acoustic noise due to gradient coils switching during scanning [30], [39]–[42].
2) *Seismocardiography (SCG)* uses accelerometer or gyroscope for measuring acceleration of the thoracic wall or vibrations of chest surface caused by heart contraction and blood ejection. The method of SCG measuring is akin to PCG method, including the necessity of filtration of gradient switching noise. The triggering precision of SCG has been confirmed in 3-T MR [43] and also in computed tomography and nuclear medicine studies [44], [45].
3) *Doppler Ultrasound (DUS)* reflects the cardiac wall motion and blood flow and can provide more detailed information about distinct points in cardiac cycle than ECG. Triggering using DUS was successfully tested in MR field up to 7 T. However, this technique fails in the case of some disorders, such as tachycardia, mitral valve insufficiency, or myocardial infarction [46]–[49].
4) *Self-Gating* methods receive the triggering information directly from MR signal by calculating changes or movements in the image volume of the investigated organ (i.e., heart or diaphragm). Despite these methods being accurate in healthy persons and being comfortable due to having no required sensors attached to the body, their correctness decreases when high HR and some heart disorders manifested by mild contraction and relaxation of myocardium are present. Moreover, the complexity and cost of self-gating software could put this method out of the reach for some MRI facilities [50]–[53].

Despite having relatively higher accuracy, among the alternative approaches that were described above which require use of physical sensors have two main drawbacks: 1) *lack of standardization for placement of the required sensors* which results in different signal morphology and consequently different outcome when placed on different sites of the body and 2) *significant signal delays* compared with ECG (see Fig. 3), which are inherent in the sensing principle of the method and in signal transmission or filtration [19]. Therefore, it is advantageous to develop measuring systems without the need for attachment of sensors to patient's body. One dominant trend in the recent years has been to make these sensors easier to apply by making them or embedding them into textile, matrasses, or cushions [54]. As will be described below, this trend is also suitable for vital sign monitoring using FOSs.

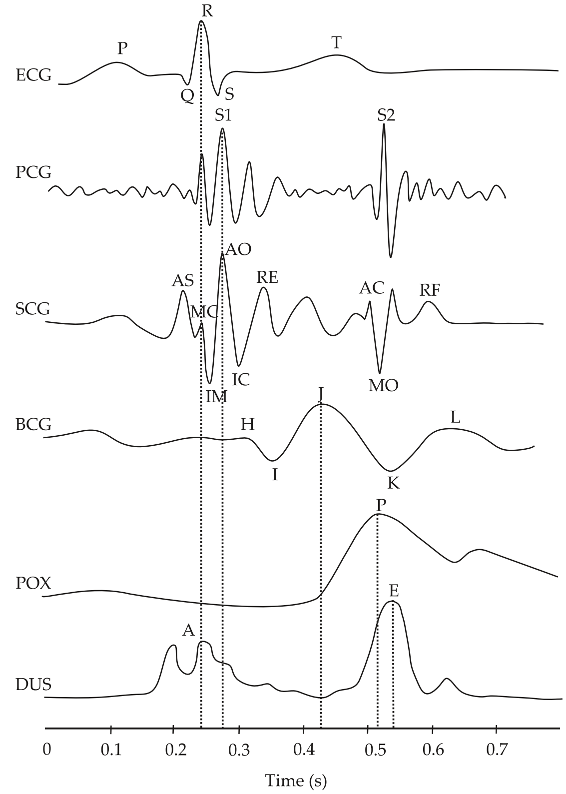

Fig. 3. Example of cardiac signals used by alternative monitoring methods and the illustration of their delay compared with the ECG signal.

C. Fiber-Optic Approach

The advent and evaluation of FOSs in multiple studies have established that they provide a promising alternative approach to measuring cardiorespiratory activity in many health fields and, in particular, for MRI applications. Despite many disturbances occurring in MRI environment, FOSs provide undistorted signals suitable for peak detection and classification of cardiorespiratory cycle phases, which are needed in MRI triggering.

Based on the type of FOS used, the sensor enables one to measure respiratory activity alone or in combination with cardiac activity, which is usually superimposed to the sensed respiratory signal (see Fig. 4). This possibility promises the elimination of sensors attached to the patient body and associated conductive wires. The individual signals that can be acquired from FOS signal are as follows.

1) *Respiratory Signal*—Its extraction consists of a prefiltering phase, which removes low-frequency fluctuations and high-frequency noise (e.g., 0.5–2 Hz) and a detection phase. The detection phase comprises pinpointing the local maxima and computing the RR. Furthermore, the inspiration and expiration phases of respiratory cycle can be obtained [55], [56].

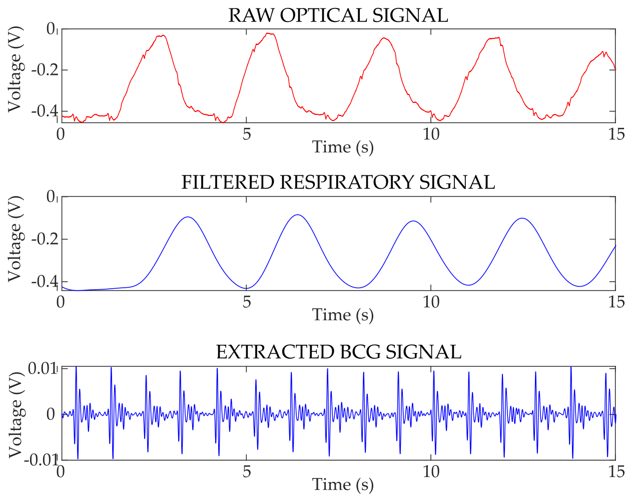

Fig. 4. Example of raw signal obtained by FOS (cardiorespiratory activity and noise) and extracted breathing and cardiac curves by applying bandpass filters (cutoff frequencies 0.5–2 Hz and 2–60 Hz, respectively).

2) *Ballistocardiography (BCG) Signal*—Represents the mechanical activity of the heart and blood ejection from the heart to the bloodstream. HR detection from optical signal is more challenging than of RR since the BCG signal must be separated by appropriate filtering of raw optical signal, typically in the range of 2–60 Hz. From the filtered signal, significant peaks (usually the most prominent J-waves) are detected [56]–[58].

III. Fiber-Optic Sensors in MRI

To explore the advantages of FOSs, it is helpful to first consider the fundamentals of their design and fabrication and then examine their application for monitoring cardiorespiratory activity. In this section, the structure of more prominent FOSs is initially described and then followed by a review of their applications.

A. Types of Sensors

Many types of FOSs have been designed and are classified by their placement sites on the patient's body (e.g., nostrils, head, back, or thorax) as well as by the principle of their operation (i.e., measuring humidity, temperature, force, elongation, strain, or pressure). The different types of FOSs that so far have been used in cardiorespiratory monitoring during MRI are summarized in the following (see Fig. 5) [10], [15].

1) *Intensity-Modulated Sensors* are among the first type of FOSs that have been investigated. Their function is based on sensing the level of humidity in humidified air from nose or mouth, which affects the interaction between the light from the optical fiber and the surrounding medium. The respiratory curve can be obtained since dry cool air does not produce condensation during inspiration, and the increased amount of light is reflected into the fiber [59]. In contrast, during expiration, the air condensates and causes the decrease in the amount of reflected light due to the water film that forms on the optical fiber [60].
2) *Microbending and Macrobending Sensors* are designed according to a general principle which is based on sensing the light transmission loss caused by deflections of an optical fiber, which results from body movements, such as cardiorespiratory activity. Another characteristic of this subgroup of FOSs is the ease of implementation through the use of measuring pads which are fabricated by sandwiching the sensor between appropriate materials within the pad (see Fig. 5). Alternatively, they have also been embedded into suitable textile [16], [17], often referred to as smart textiles. These implementations take advantage of sensing the stretching of the fiber optic, relatively to the variation in the substrate length, to detect light intensity changes.
3) *Polymer (Plastic) Sensors* are a variant of the microbending sensors which measure the intensity of a light beam shone onto a plastic part, e.g., polymethyl methacrylate (PMMA) optical fiber tip [61]. In these FOSs, different types of polymers are used, depending on the parameter that is being sensed (e.g., movements, humidity, or temperature) [61]–[63]. For sensing movement such as respiratory activity, the light beam is typically reflected from a mirror surface that its position is actuated via a spring according to the respiratory rhythm [15]. Temperature variations that accompany the respiratory cycle in mouth and nose can be measured using thermochromic pigment and temperature-sensing film that modulate the intensity of the reflected light by their color variation.
4) *Fiber Bragg grating (FBG)* sensors consist of a periodic perturbation of the refractive index along the fiber core length obtained by exposure of the core to an intense optical interference pattern. An FBG can be considered as a short segment of a fiber optic that is reflecting a narrow range of wavelengths (Bragg wavelength λ_B) and transmitting all others. The wavelength of the reflected input light is sensitive to temperature and strain. Thus, the FBG sensors can be used for measuring temperature, pressure, flow rate, and vibrations [64], [65]. Their application in medical field is advantageous mainly due to their small size, excellent measurement properties, possibilities of multiplexing [66], and lack of interference from electromagnetic and acoustic noise. Also, FBG sensors are relatively resistant to mild movements (gentle movements of the legs, hands, rapid breathing, coughing) which are often difficult to control by the patient, but the sensors are responsive to more vigorous motion [67]. Therefore, usually no compensation of motion artifacts needs to be introduced during routine MR examination. However, in exercise MRI, the signal quality may not be satisfactory.

B. Evaluation Metrics

The quality evaluation of cardiorespiratory monitoring and MRI gating using FOSs can be divided into two main approaches, i.e., signal quality evaluation and image quality assessment. Both can be performed subjectively (visual evaluation of signal morphology, amount of artifacts or noise present in signal/image, diagnostic quality of image) or objectively by means of evaluation metrics usually involving the reference methodology. In this section, the methods most often used for quality assessment in the presented publications are described.

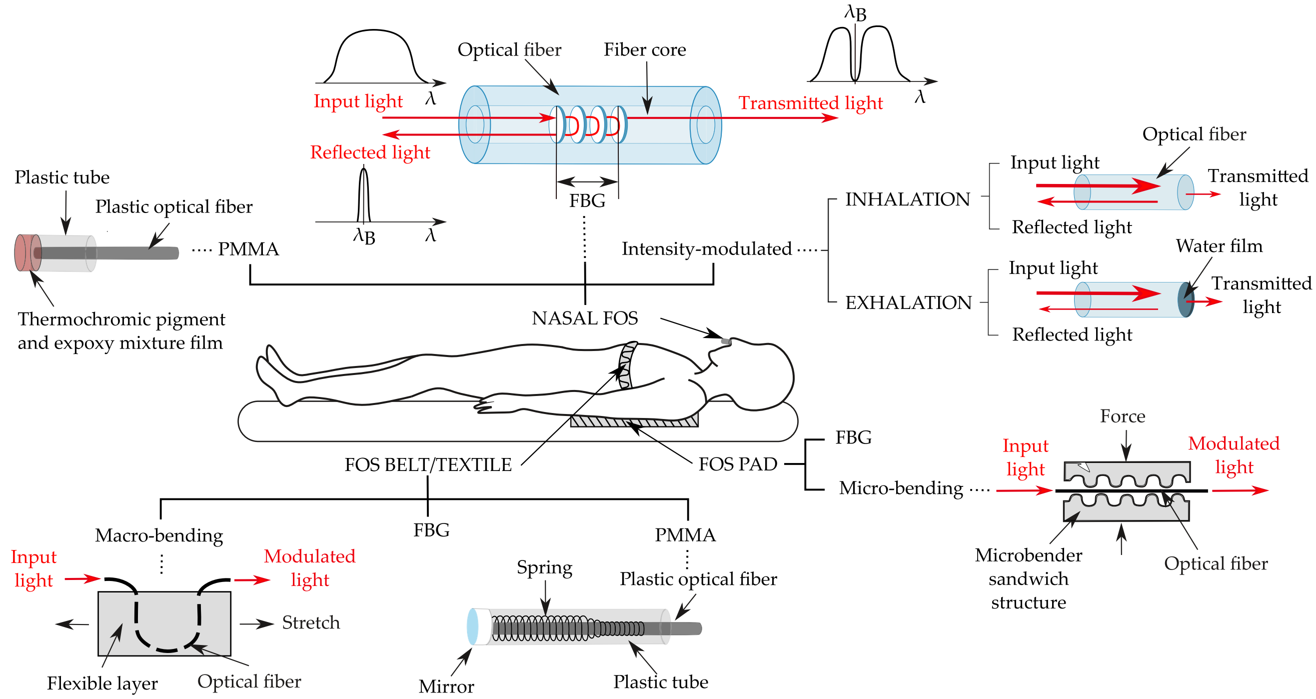

Fig. 5. Summary of principles of FOSs used in MR environment and their applications.

The signal quality and peak detection precision can be evaluated using the following parameters.

1) *Correlation Coefficient* reflects the relationship between the optical signal and the reference one. The value of correlation coefficient ranges from 0 to 1 and shows the similarity of the shape of these two signals—a higher value means a lower difference between the signals' morphology [59].
2) *Bland–Altman Analysis* is used for statistical evaluation of the difference between two signals or vectors of detected RR/HR obtained from the optical signal and the reference one. From the vector of differences, the limits of agreements (LoA) are determined as $\mu \pm 1.96\sigma$ (standard deviation). The number of samples lying within LoA determines the precision of peak detection regarding [68].
3) *Relative Error* indicates the degree of difference between optical and reference signals or peak detection. It is given as the ratio of the absolute error of optical and reference measurement to reference measurement. The closer the relative error is to zero, the more accurate the detection is.

The image quality can be assessed by means of the following methods.

1) *SNR* represents the ratio between useful signal and noise in decibels (dB). In MR images, the ratio is calculated from signal intensities of two different regions of interest (ROIs) in the same image, when the mean of signal intensities in ROI is divided by the standard deviation of noise from the surrounding [69].
2) *Contrast-to-Noise Ratio (CNR)* determines the difference in signal intensities between two ROIs divided by the standard deviation of noise from the background of the image [69].
3) *Nonreference Quality Evaluators* include Blind/Referenceless Image Spatial Quality Evaluator (BRISQUE), naturalness image quality evaluator (NIQE), and perception-based image quality evaluator (PIQE). These algorithms can evaluate the distortion of the image using natural scene statistic models. The lower the value of these parameters, the better the quality of the evaluated image [70].
4) *Structural Similarity Index Measure (SSIM)* provides information about image degradation as a structural change by means of calculation of dependencies of individual pixels. The calculation of similarity is performed by comparison of brightness, contrast, and image structures. The value of SSIM ranges from -1 to $+1$, when $+1$ is achieved when the compared images are identical [69].
5) *Subjective Evaluation* is performed by radiologists or related clinicians using blind questionnaire. The three- to five-point scale system is usually used. The experts assess the overall image quality, recognizability of the individual structures or clinical symptoms, image sharpness and artifacts' occurrence, and stage of the heart cycle in which the scan is obtained.

C. Respiratory Monitoring

For respiratory monitoring during MRI, various types of FOSs have been investigated and tested. They are designed with the goal of easy and universal application. Specifically, they can be embedded into pads, harnesses, or easy-to-use nostril sensors. These solutions could facilitate and expedite MR examinations and also reduce the patient's stress from protracted preparations for examination. The studies dealing with these sensors are classified in Section III-C and summarized in Table II.

TABLE II
SUMMARY OF STUDIES ON RESPIRATORY MONITORING USING FOSs IN MRI

Ref	Sensing element, placement	Subject, sequences, results	MRI	Triggering
[71]	Intensity-modulated FOS, nostril	289 patients, T2-weighted fast spin-echo (FSE), method rated as valuable in 94 cases	0.2 – 1.5T	No
[59]	Micro-bending, mat under back	20 healthy subjects, T2-weighted half-Fourier single-shot turbo spin-echo (HASTE), correlation r = 0.971, no significant difference in the SNR and CNR between FOS and navigator-gated MR images, significant difference in subjective diagnostic quality	1.5T	Yes
[63], [74]	Thermochromic pigment and spring sensors, nasal cavity and abdomen	Phantom MR images, no information about subjects, spin-echo (SE) T1-weighted and gradient-echo (GE) T2-weighted sequences, no results presented	0.32T	No
[79], [76] [80], [81]	OFSETH – combined macro-bending and FBG, chest and abdominal harness	Phantom and in vivo MR images, no additional information presented	3T	No
[37]	FBG in nasal cannula, nostril	10 healthy subjects (6 in laboratory conditions, 4 in MRI), T2-weighted turbo spin-echo (TSE), RR total relative error < 5 %, higher image quality according to both subjective and objective analysis (BRISQUE)	3T	Yes

1) Nasal Sensors: Larsson *et al.* [71] proved the viability of respiratory monitoring using FOSs during MRI for three different MR devices with the field strength ranging from 0.2 to 1.5 T. The sensor, working on the principle of sensing the changes in condensed humidity during respiration, is placed in front of one nostril. In this way, it senses expired air from both the nose and mouth. Its accuracy was proven in laboratory conditions compared with reference acoustic method and transthoracic impedance plethysmography in 1994 [72]. Later, the precision of this approach was compared with manual and capnography method, and it was shown that the fiber-optic monitor provides comparable results for RR detection. Complications occurred in patients with nasal septum deviations that cause turbulence in air stream, leading to measurement errors [73]. For MR compatibility verification, a total of 289 patients were examined, encompassing different types of examinations such as spine, brain, and orthopedic. Respiratory monitoring using the proposed method was evaluated via the level of its necessity for examination. In this study, the number of patients with complicating conditions, such as claustrophobia, anxiety, pain, sedation, epilepsy, and nausea, was 75. In 52 of these patients, respiratory monitoring was deemed to be necessary/valuable and not necessary in the remaining 23 cases. In the case of other 214 patients without complicating conditions, the method was deemed to be necessary/valuable in 42 cases.

Yoo *et al.* [63] presented two types of FOSs for measuring the respiratory activity during MRI. The first one was placed in the nasal cavity and included a thermochromic pigment that changed its color according to the temperature variation in the respiratory air flow. Specifically, it sensed the higher temperature during expiration with pigment color of white and lower temperature during inspiration with the pigment color of red. The second sensor measured abdominal circumference changes using PMMA tubes, a mirror, and a spring. The respiratory-driven abdominal movements were transduced to the mirror movement via the connecting spring, and the respiration activity was sensed by measuring the variation in the distance between the mirror and the distal end of the plastic optical fiber. These signals were compared with conventional respiratory sensors (temperature sensors and respiratory belt). The study was performed using a water phantom in 0.32-T MRI environment. The first FOS sensor that was placed in the nasal cavity was visible in MR images, while the abdominal FOS sensor was not. Hence, the abdominal sensors are considered to be suitable for use during MRI, without causing any artifacts. Also, the authors proved that the signals from both the sensors were not distorted by magnetic field or electromagnetic interference and had high sensitivity and desirable SNR.

Yoo *et al.* [74] proposed another type of nasal cavity sensor. The sensor measures the variations in infrared radiation generated by the respiratory airflow from a nasal cavity using a silver halide optical fiber. The infrared radiation occurs during expiration, when the nasal tissue is warm. The silver halide FOS was shielded from the effects of magnetic fields and RF pulses using a box that was made of copper and aluminum. The results were compared with the previous FOS designs that used temperature-sensing film and the conventional sensor with temperature transducers. Furthermore, phantom and *in vivo* brain MR images were acquired under the same conditions as in the previous study (i.e., 2010 study). The distal sensing probes of the novel FOS design were again visible in MR images, and the respiration curves were comparable for all the investigated sensors without any distortion caused by MR interference.

Fajkus *et al.* [37] introduced a novel breathing sensor that was based on embedding FBG into conventional nasal oxygen cannulas. The sensor was tested under laboratory conditions and in clinical 3-T MRI environment. The performance was evaluated related to the standard pneumatic respiratory reference using the Bland–Altman analysis. The quality of triggered images was evaluated by radiology experts and BRISQUE [75]. The acquired images were found to be comparable and even higher quality than those obtained using standard respiratory belt triggering method, providing sharp contours, high contrast of desired structures, and no motion artifacts.

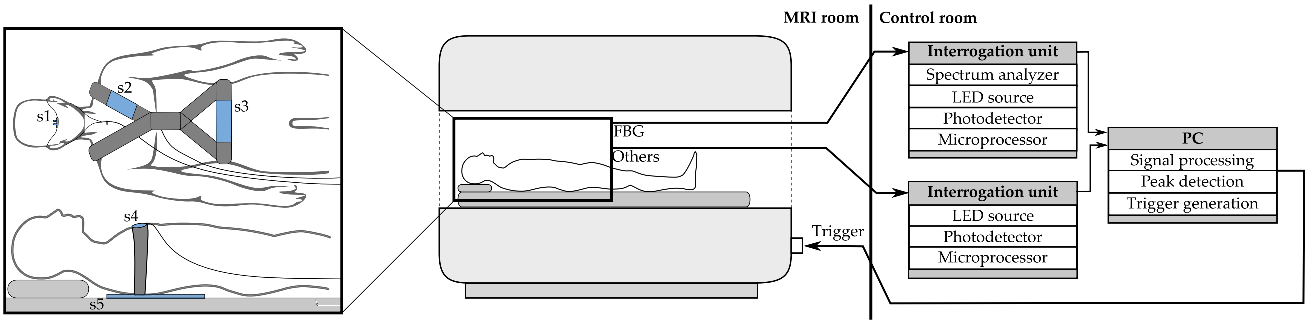

Fig. 6. Illustration of the experimental setup with different FOSs used: nasal sensor (s1—e.g., PMMA [63] or FBG [37]), sensors within harness (s2—FBG, s3—macro-bending [76]), chest sensor (s4—e.g., FBG encapsulated in PDMS [70] or fiberglass [77]), and pad sensor (s5, e.g., microbending pad [78] or FBG springboard [68]).

TABLE III

SUMMARY OF STUDIES ON CARDIAC MONITORING USING FOSs IN MRI

Ref	Sensing element, placement	Subject, sequences, results	MRI	Triggering
[58], [82]	FBG on plexiglass board, under back	3 subjects, T1-weighted SE and T2-weighted GE, HR samples within LoA > 95 %, HR maximum relative error span 6.61 %	1.5T	No
[70]	FBG in PDMS, chest	8 subjects, Free Induction Decay Steady-State Precession (True FISP) and Phase Sensitive Inversion Recovery (PSIR), comparable image quality with standard ECG triggering (BRISQUE, NIQE, PIQE, subjective evaluation)	3T	Yes
[69]	FBG in PDMS, chest	10 subjects, TRUEFISP and PSIR sequences, comparable image quality with standard ECG triggering (SNR, CNR, BRISQUE, NIQE, SSIM, subjective evaluation), reduced time of examination	3T	Yes

2) Harness Sensors: In 2008, the project involving optical fiber sensors embedded into technical textile for healthcare monitoring (OFSETH) was first introduced [79]. The smart textile in the form of elastic harness was designed for monitoring respiratory activity using two types of FOS—FBG (thoracic area) and macro-bending (abdominal area) sensors (see Fig. 6). The project focused on monitoring during MRI, so emphasized the MR-compatible materials and signals not affected by MR interferences.

Witt *et al.* [76] proposed sensing harness for MRI applications based on silica and polymer fiber, implemented into textile that can sense elongation of up to 3%, while maintaining the stretching properties of the textile substrates for patient comfort. The harness is designed for measuring respiratory activity from both abdominal and thoracic areas using macro-bending principle, FBG, and time-domain reflectometry. The proposed solution was also tested in 3-T MRI environment [80]. Safety issues were considered in this study, and MRI compatibility was proven using a phantom and a simulator.

Furthermore, in 2011, Witt *et al.* [81] tested the sensory harness on a healthy subject. In this study, a new duplex optical connector was introduced to reduce the number of connections and optical fibers. The optical connector's core included two ceramic ferrules integrated into a plastic housing, making the whole connector MRI-compatible. The signal was compared with the one obtained by a spirometer. In MRI environment, the sensors were examined on the simulator. The signal obtained by the proposed sensor was not degraded by the gradient field of MR and showed the typical patterns in both thoracic and abdominal respiratory movement.

3) Pad Sensors: Lau *et al.* [59] described novel microbending fiber-optic system for respiratory monitoring and MRI gating. The FOSs were embedded into a mat that was placed on the MRI bed, under the patient's diaphragm region. The sensor design was created as a "sandwich" structure, which deforms the optical fiber inside according to the mechanical perturbations (respiratory or body motion) and induces the signal presenting this motion. The sensor system was validated using 1.5-T MRI environment. The reliability of RR detection was compared with the conventional pneumatic respiratory bellow, placed around the subject's lower chest. A significant correlation ($r = 0.971$) in RR was shown between FOSs and respiratory bellows. For MRI gating, single-slice liver images were obtained and compared with triggering using navigation echoes. The image quality was assessed according to the overall image quality, the clarity of liver and hepatobiliary features, and the presence of artifacts. No significant difference in the liver SNR and liver–spleen CNR between FOS and navigator-gated MR images was found. However, these images showed a significant difference in diagnostic quality. In about 40% of cases, FOS-gated images reached comparable or even better quality than the navigator-gated ones, and only rare occasions of artifacts or unreadable data occurred.

D. Cardiac Monitoring

For cardiac monitoring, FBG sensors are advantageous, thanks to their high sensitivity, precision, and small dimensions, as reported in the literature presented here. The studies dealing with FOS cardiac monitoring are described below and summarized in Table III.

Dziuda *et al.* [82] described the FBG-based sensor for measuring heart activity mounted on a spring plate that was placed between patient's back and the bed mattress. This board transmits body vibrations to FBG, reflecting function of lungs and heart. The prototype also contained a Plexiglass plate, where FBG was attached using epoxy adhesive. The authors presented possibilities of measuring both respiratory and heart activities, which can be extracted from the optical signal using filtration. The quality of detection of HR was determined using the Bland–Altman analysis in laboratory conditions and using 1.5-T MRI compared with ECG with promising results.

Dziuda and Skibniewski [58] proved MRI compatibility and resistance to MR-related artifacts of FBG sensor in the form of an elastic board that was placed under patient's back. Two versions of FBG sensors encompassing two different sizes of PMMA boards where the sensors were placed were tested. Before testing the sensor in MRI environment, the experiments were conducted under laboratory conditions, comparing signals obtained from standing, sitting, and supine positions. In MR environment, the quality of HR detection was evaluated by comparing it to standard ECG measurement, and the maximum relative error span of 6.61% was achieved. The study proved that the presence of the sensor is safe for the patient during MRI and has no effect on image quality.

Nedoma *et al.* [70] performed cardiac triggering using the FBG sensor prototype with polydimethylsiloxane polymer (PDMS) encapsulation and in this time placed on the chest. In this study, optimal sensor placement was investigated first in laboratory conditions to generate greater BCG signal amplitude and it was found to be dependent on the size of the contact area of the sensor with the patient's body. The authors found that the sensor should be placed on the chest above the site of mitral or tricuspid valve, where the signal reaches the peak in the ventriculi during systole. The experiments were conducted using 3-T MRI on eight subjects. For cardiac triggered examination, cinematic T1/T2 balanced and phase-sensitive inversion recover sequences were used. Since the accuracy of HR and RR detection was proven earlier [67], only the triggered image quality was evaluated by both objective and subjective analyses, based on BRISQUE, NIQE, and PIQE algorithms and blind questionnaire, where ten experts evaluated the diagnostic quality of images. The results were compared with ECG and POX-triggered images. According to the results of objective analysis, the parameters reached comparable values in all three methods. The subjective analysis showed the comparable diagnostic value of the FOS-triggered images, which outperformed the POX-triggered image quality. The ECG outperformed FOS, but there were no significant differences (average score of subjective evaluation was 1.55 for ECG and 1.7 for FOS).

Brablik *et al.* [69] evaluated the suitability of FOS cardiac triggering regarding not only ECG but also another BCG approach based on the pneumatic sensory system. The previously described prototype of FOS [70] was used during true fast imaging with steady-state precession (TRUEFISP) and phase-sensitive inversion recovery (PSIR) cardiac sequences. The quality of triggered images was evaluated using SNR, CNR, BRISQUE/NIQE methods, and subjective analysis by radiologists. According to these parameters, no significant differences were obtained between individual approaches. In addition, the time needed for examination was measured, including patient's preparation. The study proved that alternative approaches can overcome the ECG method in terms of application of sensors to the patient and also in the effectiveness of acquisition triggering. Moreover, comfort rating of the tested subjects showed better results in the case of pneumatic and fiber-optic methods.

E. Combined Monitoring

Several FOS prototypes enable monitoring of both cardiac and respiratory signals simultaneously, which reduces the number of wires and measuring units that need to be applied to the patient. Moreover, they enable combined MRI triggering that aims to reduce the examination time, increase patient's comfort, and provide high-quality images without artifacts caused by cardiorespiratory movements. The studies on these approaches are described below and summarized in Table IV.

Dziuda *et al.* [68] introduced new study of their latest sensor, also comparing the respiratory detection with pneumatic bellows placed on patient's chest (12 subjects). Each measurement lasted 5 min, and the mean relative error reached $5.24\%\pm1.02\%$ and $5.71\%\pm0.61\%$ for of RR detection and HR detection, respectively. The study also reported the subjects' responses about any discomfort or other negative feelings as well as those cases where there were no complaints. Electromagnetic fields and other MR interferences did not affect the signal quality, as the quality was equal to that obtained without MRI exposure. Also, the images were not distorted by any component of fiber-optic measuring system.

Another study was conducted by Dziuda *et al.* [83] during thoracic spine examinations in 1.5-T MRI scanner, this time collecting data (three patients) through complete MR examination (from 18 to 39 min). The Bland–Altman analysis showed a sufficient performance of RR and HR detection using FBG sensor with maximum relative errors of 7.67% and 6.61%, respectively. The authors discussed the effect of motion artifacts that were visible in FOS signal, but such artifacts did not affect proper peak detection since they were of short duration and patients were asked to lie still during examination.

Chen *et al.* [78] proposed a highly sensitive microbending FOS for measurement of breathing rate and HR. They embedded the sensor into a sensory mat that wirelessly communicated using a battery-powered transceiver with a computer via Bluetooth. The individual vital functions were extracted from the whole optical signal by digital signal processing algorithm, including noise suppression, filtering (for HR extraction), signal averaging (for RR extraction), and peak detection. The study was performed in 3-T MRI environment on 11 subjects, and FOS signals were compared with the use of commercial respiratory bellows and pulse oximeter. The computed correlation coefficients were 0.963 and 0.997 for RR and HR, respectively. The sensors did not cause any distortion in the acquired images, nor they were affected by radio frequency artifacts. The authors stated that their peak detection algorithm is independent of the irregular heart beating or breathing excursion and can be applied for abnormal breathing and heartbeating subjects.

TABLE IV

SUMMARY OF STUDIES ON COMBINED MONITORING USING FOSs IN MRI

Ref	Sensing element, placement	Subject, sequences, results	MRI	Triggering
[78]	Micro-bending, mat under back	11 subjects, sequence not specified, RR correlation r = 0.963, HR correlation r = 0.997, RR relative error $<$ 10 % (absolute error $<$ 2 bpm), HR relative error $<$ 1 % (absolute error $<$ 0.5 %)	3T	No
[68], [83]	FBG springboard, under back	3 – 12 adult subjects, T1-weighted SE and T2-weigtened GE, RR maximum error $\pm$1.2 bpm (relative error 7.67 %), HR maximum error $\pm$3 bpm (relative error 6.61 %), HR samples within LoA $>$ 94.88 %, RR samples within LoA $>$ 95.38 %	1.5T	No
[84]	FBG T-shirt, chest and abdomen	2 healthy subjects, T1-weighted and T2-weighted SE and FSE sequences, frequency analysis,no results presented	1.5T	No
[86]	FBG with PDMS, under back	4 subjects, sequences not specified, HR maximum relative error 5.86 %, mean HR samples within LoA 95.49 %, RR maximum relative error 4.41 %, mean RR samples within LoA 96.1 %	1.5T	No
[77]	FBG in fiberglass, chest	10 subjects, T1-weighted SE and T2-weigtened GE, Diffusion Weighted Imaging (DWI), RR and HR samples within LoA $>$ 95 %, RR error $\pm$0.8 bpm, HR error $\pm$3.6 bpm, HR and RR relative error $<$ 5 %	1.5T	No

Presti *et al.* [84] described wearable smart textile in the form of T-shirt for monitoring both heart and respiratory activities. The system consists of six FBG sensors and was tested in conjunction with the use of 1.5-T MRI. In this study, the frequency responses of the measured signals were analyzed to evaluate the sensitivity of the sensor and influence of the interference from environment. The performance of the proposed system was observed to match that of a previous study that was performed under laboratory conditions [85] in which the RR maximum error reached 1.14%.

In the same year, Nedoma *et al.* [86] validated a novel FBG sensor for simultaneous monitoring of cardiac and respiratory activities. The study was conducted using a 1.5-T MRI with four subjects. The FBG was encapsulated inside a PDMS which is inert and does not react with human skin and electromagnetic field; the sensor was placed underneath the patient's back. Also, the sensor did not induce any artifacts into images. The results were evaluated using the Bland–Altman analysis with reference ECG and signal from respiratory belts. The maximum relative errors reached satisfactory levels of 4.41% for RR and 5.86% for HR analyses.

Another prototype of the sensor was described in 2019 [77], where the FBG was encapsulated into fiberglass. This solution made it possible for the sensor to have very small dimensions (30 $\times$ 10 $\times$ 0.8 mm) and weight (2 g) and it was attached to the patient's chest using elastic belt. Cardiorespiratory monitoring was performed using 1.5-T environment with ten subjects. The sensor was not visible in MR images; therefore, it did not affect image quality. The signals were compared with reference ECG and pneumatic elastic respiratory transducer which are commonly used in MR examinations using the Bland–Altman analysis. The maximum relative errors reached 4.94% for RR and 5.68% for HR detection. For the first time, the authors discussed and presented the possibility of cardiac triggering of MR acquisition and possible challenges of time delay of the BCG signal compared with ECG.

IV. DISCUSSION

Since FOS technology brings many advantages that can be used in healthcare, many authors have dealt with its implementation into novel sensory systems. These innovations could, in the future, replace some of the current conventional methods in monitoring during MRI. These systems are resistant to electromagnetic and other interferences that pose many challenges to physiological activity monitoring via standard techniques that are currently used in clinical practice, such as ECG.

The popularity of investigating FOSs in MR environment started to increase ten years ago, as shown in Fig. 7. After the first attempts with humidity sensors, the research started to focus mainly on FBG and micro/macro-bending sensors which outperformed the humidity and temperature monitoring sensors, specifically because of their ease of use. The literature survey showed that FBG sensors are the most often investigated type of FOSs for MRI applications due to their high sensitivity and the possibility of combining the measurements of both cardiac and respiratory activities. Most experiments proved that the RR and HR detection accuracies with FOSs have reached greater than 95% compared with current standard methods and attained comparable triggered image quality.

Depending on the sensor design, FBG sensors can be implemented as simple chest sensors attached by an elastic band or embedded into a pad and smart textiles, such as harness and T-shirt. Such sensors can provide a quick and easy-to-use solution when the patient's preparation is minimized in contrast to the use of electrodes that often requires cleaning and shaving the skin. Therefore, the application of FOS technology could enhance the workflow and patient comfort, which eliminates the additional stress on both the patient and medical staff. However, this kind of implementation is challenging, because it requires arriving at an optimal size and shape of the smart textile that would be applicable to all the patients regardless of their anatomy. However, at times, it may also be challenging to ensure proper contact of the sensor with the

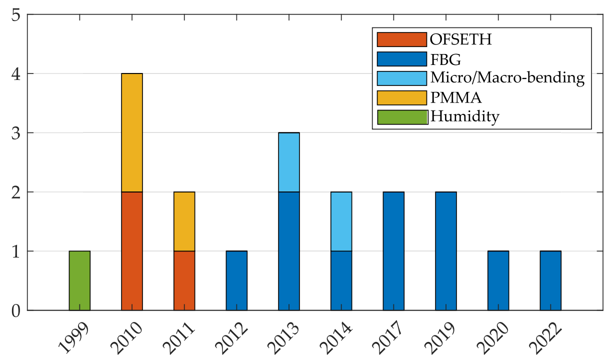

Fig. 7. Number of publications over years on FOS experiments in MRI environment.

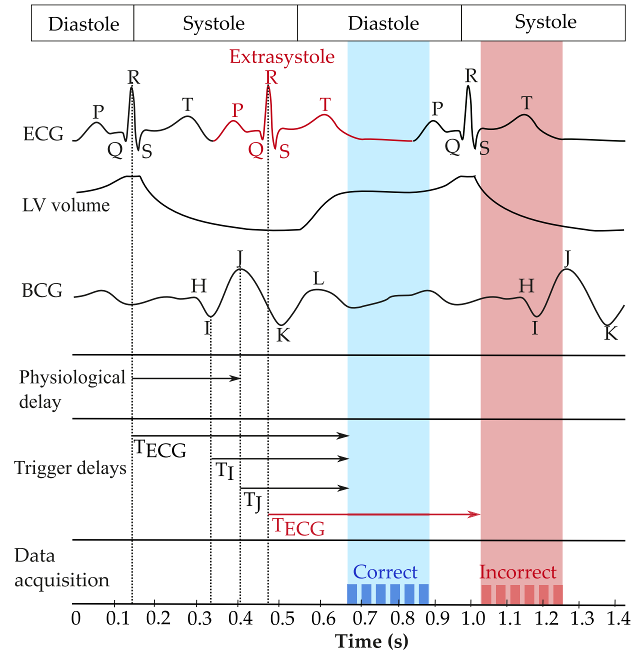

Fig. 8. Example illustration of heart trigger timing using BCG and ECG regarding left ventricular (LV) volume when imaging diastolic phase and occurrence of extrasystole (red). Blue—proper and red—wrong data acquisition timing.

patient's body and its accurate positioning by some universal prototype. Therefore, the development of simple sensor for placement on the patient's chest or underneath his/her back still remains a significant part of the research.

An indisputable advantage of FOS is its reusability in comparison to the disposable ECG electrodes, making it low-cost and environmentally more friendly. However, the price of the FOS-based measuring systems must be discussed from many aspects. While the sensor is a low-cost alternative to current technologies that are in use in clinical practice, the measuring unit (especially for FBG-based systems) is a quite costly device, causing the system's price to be comparable with the conventional ones. On the other hand, the measuring unit needed for conversion of the optical signal can usually be used in multichannel mode and thus can process signals from several sensors, i.e., several MRI devices/examination rooms.

Regarding the quality of vital sign monitoring and MRI triggering by FOSs, several observations can be made.

1) *Delay of the BCG Signal* that is present when detecting cardiac activity (see Fig. 8) can cause problems with scanning the desired phase of cardiac cycle. The physiological delay of BCG signal is about 200 ms (the mean from alternative methods; PCG/SCG ~30 ms, DUS ~400 ms) [19], [87]. Some level of delay after occurrence of a significant peak can render the triggering system unusable, as it would be too late for starting data acquisition and completely capturing the desired phase. Of course, such phenomenon can result in the origin of artifacts and decreased image quality, even the loss of diagnostic information [24], [88]. Therefore, the compensation of delay should be performed. The simplest way to do that is the customization of trigger delay, which is shortened compared with ECG trigger delay. Another way of delay compensation is the detection of earlier negative I-wave as a start point of the cycle. In the case of retrospective gating, the delay compensation must be introduced into the reconstruction software.
2) *Mechanical Heart Activity* monitoring versus electrical activity brings advantageous properties. In addition to the possible advantages that this monitoring method offers with respect to the patient's preparation and electrodes' safety precautions mentioned earlier, it is important to highlight the limited accuracy of ECG in the case of extrasystoles. That is, the occasions in ECG signal that are caused by abnormal function of the cardiac conduction system [89]. In such a case, the system considers the captured R wave as the activator of the next cardiac cycle, so the calculation of HR may be distorted, and triggered data acquisition may occur in the wrong part of cardiac cycle (see Fig. 8). Nowadays, several MR systems are partially immune to the extrasystoles, thanks to incorporating a triggering window in which it anticipates the next R wave [24]. However, it can still be hard to set proper triggering parameters (such as target R–R interval or triggering delay) due to incorrect HR detection. In contrast, BCG-based triggering does not suffer from these errors since the extrasystoles do not result in the mechanical activity of the heart, and thus, only actual heart contractions are detected [90].
3) *Optimal Placement of Sensors* depends on the purpose of monitoring. As one can see from the presented literature survey, several main locations that are used with FOSs are nostril, chest, back, or a combination of chest and abdomen. The implementation of nasal prototypes [37], [63], [71], [74] seems to be the easiest and most convenient approach due to its ease of application and universality. However, this sensor type enables just respiratory monitoring that is insufficient for many applications when pulse oximeter or ECG needs to be used as well. Therefore, sensor prototypes allowing heart monitoring are involved in many studies—placed under back [58], [59], [78], [82], [83], [86] or on chest [68], [70], [77], [80]. Although this approach produces high-accuracy measurement of HR and RR, optimal sensor placement

is not described in any study. The reason is that the morphology of BCG signal varies depending on the patient's physiology and anatomy [91], [92]. Thus, the finding of optimal placement is quite a challenging task, which needs to be addressed by design of more universal measuring pads or textiles.

V. Conclusion

This article provides an insight into measuring the cardiorespiratory activity for high-quality MR image acquisition using FOSs. It briefly describes the principles of the sensors used for this purpose and summarizes recent studies performed on this topic. The proposed solutions appear to be viable alternatives to conventional methods of cardiorespiratory monitoring and MRI triggering, due to their compatibility with MRI, ease of use, immunity to MR interference, and reusability of the sensors. Another important benefit of sensing mechanical signals is immunity to extrasystoles, which usually cause false acquisition triggering. However, many challenges remain to be solved for successful implementation of this new approach to clinical practice, such as delay of BCG signal that can decrease triggering accuracy or optimal placement of the sensor, when improper positioning can decrease the signal quality (motion artifacts, extreme sensitivity to noise, reduced signal amplitude, etc.). In connection with the above, the complete monitoring system suitable for deployment in clinics has yet to be designed to provide more satisfactory results. Furthermore, the presented studies dealt with preliminary prototypes of FOSs, and most of the investigators have reported on studies involving a relatively small number of subjects. The future research would include the industrial design of the sensor prototype that would be suitable for more extensive and detailed clinical studies, which should be performed to validate all the benefits of this alternative approach.

References

[1] M. E. Ladd *et al.*, "Pros and cons of ultra-high-field MRI/MRS for human application," *Prog. Nucl. Magn. Reson. Spectrosc.*, vol. 109, pp. 1–50, Dec. 2018.

[2] A. Stadler, W. Schima, A. Ba-Ssalamah, J. Kettenbach, and E. Eisenhuber, "Artifacts in body MR imaging: Their appearance and how to eliminate them," *Eur. Radiol.*, vol. 17, no. 5, pp. 1242–1255, Dec. 2006.

[3] E. Kanal and F. G. Shellock, "Patient monitoring during clinical MR imaging," *Radiology*, vol. 185, no. 3, pp. 623–629, Dec. 1992.

[4] D. W. Chakeres, A. Kangarlu, H. Boudoulas, and D. C. Young, "Effect of static magnetic field exposure of up to 8 Tesla on sequential human vital sign measurements," *J. Magn. Reson. Imag.*, vol. 18, no. 3, pp. 346–352, Aug. 2003.

[5] D. Stäb, J. Roessler, K. O'Brien, C. Hamilton-Craig, and M. Barth, "ECG triggering in ultra-high field cardiovascular MRI," *Tomography*, vol. 2, no. 3, pp. 167–174, Sep. 2016.

[6] R. Shah and Y. Agrawal, "Introduction to fiber optics: Sensors for biomedical applications," *Indian J. Pharmaceutical Sci.*, vol. 73, no. 1, p. 17, 2011.

[7] J. I. Peterson and G. G. Vurek, "Fiber-optic sensors for biomedical applications," *Science*, vol. 224, no. 4645, pp. 123–127, 1984.

[8] D. Tosi, S. Poeggel, I. Iordachita, and E. Schena, "Fiber optic sensors for biomedical applications," in *Opto-Mechanical Fiber Optic Sensors*. Amsterdam, The Netherlands: Elsevier, 2018, pp. 301–333.

[9] P. Roriz, O. Frazão, A. B. Lobo-Ribeiro, J. L. Santos, and J. A. Simões, "Review of fiber-optic pressure sensors for biomedical and biomechanical applications," *J. Biomed. Opt.*, vol. 18, no. 5, pp. 050903-1–050903-18, May 2013.

[10] F. Taffoni, D. Formica, P. Saccomandi, G. Di Pino, and E. Schena, "Optical fiber-based MR-compatible sensors for medical applications: An overview," *Sensors*, vol. 13, no. 10, pp. 14105–14120, Jan. 2013.

[11] H. Su *et al.*, "Fiber-optic force sensors for MRI-guided interventions and rehabilitation: A review," *IEEE Sensors J.*, vol. 17, no. 7, pp. 1952–1963, Jan. 2017.

[12] C. Massaroni, M. Zaltieri, D. L. Presti, A. Nicolò, D. Tosi, and E. Schena, "Fiber Bragg grating sensors for cardiorespiratory monitoring: A review," *IEEE Sensors J.*, vol. 21, no. 13, pp. 14069–14080, Jul. 2021.

[13] P. Samartkit and S. Pullteap, "Fiber optic sensor applications for vital signs monitoring: A review," in *Proc. 7th Int. Elect. Eng. Congr. (iEECON)*, Mar. 2019, pp. 1–4.

[14] C. P. Mayoral *et al.*, "Fiber optic sensors for vital signs monitoring. A review of its practicality in the health field," *Biosensors*, vol. 11, no. 2, p. 58, Feb. 2021.

[15] Ł. Dziuda, "Fiber-optic sensors for monitoring patient physiological parameters: A review of applicable technologies and relevance to use during magnetic resonance imaging procedures," *J. Biomed. Opt.*, vol. 20, no. 1, p. 1, Jan. 2015.

[16] C. Massaroni, P. Saccomandi, and E. Schena, "Medical smart textiles based on fiber optic technology: An overview," *J. Funct. Biomater.*, vol. 6, pp. 204–221, Apr. 2015, doi: 10.3390/jfb6020204.

[17] Z. Gong *et al.*, "Wearable fiber optic technology based on smart textile: A review," *Materials*, vol. 12, no. 20, p. 3311, Oct. 2019.

[18] S. Donati and M. Norgia, "Self-mixing interferometry for biomedical signals sensing," *IEEE J. Sel. Topics Quantum Electron.*, vol. 20, no. 2, pp. 104–111, Mar. 2014.

[19] M. Ladrova *et al.*, "Monitoring and synchronization of cardiac and respiratory traces in magnetic resonance imaging: A review," *IEEE Rev. Biomed. Eng.*, vol. 15, pp. 200–221, 2022.

[20] J. Enders *et al.*, "Reduction of claustrophobia during magnetic resonance imaging: Methods and design of the 'CLAUSTRO' randomized controlled trial," *BMC Med. Imag.*, vol. 11, no. 1, pp. 1–15, Feb. 2011.

[21] Z. Siddiqui, P. Singh, S. Kushwaha, and R. Srivastava, "MRI and fear of confined space: A cause and effect relationship," *Int. J. Contemp. Med. Surg. Radiol.*, vol. 22, pp. 19–24, Jan. 2017.

[22] Ł. Dziuda, P. Zieliński, P. Baran, M. Krej, and L. Kopka, "A study of the relationship between the level of anxiety declared by MRI patients in the STAI questionnaire and their respiratory rate acquired by a fibre-optic sensor system," *Sci. Rep.*, vol. 9, no. 1, pp. 1–11, Mar. 2019.

[23] A. Elster, "Questions and answers in magnetic resonance imaging, *Radiology*, vol. 220, no. 2, pp. 524–524, 2001, doi: 10.1148/radiology.220.2.r01au28524.

[24] M. S. Nacif, A. Zavodni, N. Kawel, E.-Y. Choi, J. A. C. Lima, and D. A. Bluemke, "Cardiac magnetic resonance imaging and its electrocardiographs (ECG): Tips and tricks," *Int. J. Cardiovascular Imag.*, vol. 28, no. 6, pp. 1465–1475, Oct. 2011.

[25] H. Kugel *et al.*, "Hazardous situation in the MR bore: Induction in ECG leads causes fire," *Eur. Radiol.*, vol. 13, no. 4, pp. 690–694, Apr. 2003.

[26] S. Lange and Q. N. Nguyen, "Cables and electrodes can burn patients during MRI," *Nursing*, vol. 36, no. 11, pp. 11–18, Nov. 2006.

[27] T. Niendorf, D. K. Sodickson, G. A. Krombach, and J. Schulz-Menger, "Toward cardiovascular MRI at 7 T: Clinical needs, technical solutions and research promises," *Eur. Radiol.*, vol. 20, no. 12, pp. 2806–2816, Jul. 2010.

[28] A. M. Gharib, A. Elagha, and R. I. Pettigrew, "Cardiac magnetic resonance at high field: Promises and problems," *Current Problems Diagnostic Radiol.*, vol. 37, no. 2, pp. 49–56, Mar. 2008.

[29] D. Abi-Abdallah, V. Robin, A. Drochon, and O. Fokapu, "Alterations in human ECG due to the MagnetoHydroDynamic effect: A method for accurate R peak detection in the presence of high MHD artifacts," in *Proc. 29th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, Aug. 2007, pp. 1842–1845.

[30] T. Frauenrath *et al.*, "Acoustic cardiac triggering: A practical solution for synchronization and gating of cardiovascular magnetic resonance at 7 Tesla," *J. Cardiovascular Magn. Reson.*, vol. 12, no. 1, pp. 1–14, Nov. 2010.

[31] B. Sievers, M. Wiesner, N. Kiria, U. Speiser, S. Schoen, and R. H. Strasser, "Influence of the trigger technique on ventricular function measurements using 3-Tesla magnetic resonance imaging: Comparison of ECG versus pulse wave triggering," *Acta Radiologica*, vol. 52, no. 4, pp. 385–392, May 2011.

[32] W.-Y.-I. Tseng, T. G. Reese, R. M. Weisskoff, and V. J. Wedeen, "Cardiac diffusion tensor MRI *in vivo* without strain correction," *Magn. Reson. Med.*, vol. 42, no. 2, pp. 393–403, Aug. 1999.

[33] M. Miyazaki, S. Sugiura, F. Tateishi, H. Wada, Y. Kassai, and H. Abe, "Non-contrast-enhanced MR angiography using 3D ECG-synchronized half-Fourier fast spin echo," *J. Magn. Reson. Imag.*, vol. 12, no. 5, pp. 776–783, 2000.

[34] J. Krug, G. Rose, D. Stucht, G. Clifford, and J. Oster, "Limitations of VCG based gating methods in ultra high field cardiac MRI," *J. Cardiovascular Magn. Reson.*, vol. 15, no. S1, pp. 1–2, Jan. 2013.

[35] N. Spicher, M. Kukuk, S. Maderwald, and M. E. Ladd, "Initial evaluation of prospective cardiac triggering using photoplethysmography signals recorded with a video camera compared to pulse oximetry and electrocardiography at 7 T MRI," *Biomed. Eng. OnLine*, vol. 15, no. 1, pp. 1–28, Nov. 2016.

[36] C. Santelli *et al.*, "Respiratory bellows revisited for motion compensation: Preliminary experience for cardiovascular MR," *Magn. Reson. Med.*, vol. 65, no. 4, pp. 1097–1102, Nov. 2010.

[37] M. Fajkus *et al.*, "MR fully compatible and safe FBG breathing sensor: A practical solution for respiratory triggering," *IEEE Access*, vol. 7, pp. 123013–123025, 2019.

[38] A. C. Larson *et al.*, "Preliminary investigation of respiratory self-gating for free-breathing segmented cine MRI," *Magn. Reson. Med.*, vol. 53, no. 1, pp. 159–168, Jan. 2005.

[39] T. Frauenrath, T. Niendorf, and M. Kob, "Acoustic method for synchronization of magnetic resonance imaging (MRI)," *Acta Acustica United Acustica*, vol. 94, no. 1, pp. 148–155, Jan. 2008.

[40] M. Becker *et al.*, "Comparison of left ventricular function assessment using phonocardiogram- and electrocardiogram-triggered 2D SSFP CINE MR imaging at 1.5 T and 3.0 T," *Eur. Radiol.*, vol. 20, no. 6, pp. 1344–1355, Dec. 2009.

[41] K. Nassenstein *et al.*, "Cardiac MRI: Evaluation of phonocardiogram-gated cine imaging for the assessment of global und regional left ventricular function in clinical routine," *Eur. Radiol.*, vol. 22, no. 3, pp. 559–568, Sep. 2011.

[42] T. Frauenrath *et al.*, "Feasibility of cardiac gating free of interference with electro-magnetic fields at 1.5 Tesla, 3.0 Tesla and 7.0 Tesla using an MR-stethoscope," *Investigative Radiol.*, vol. 44, no. 9, pp. 539–547, Sep. 2009.

[43] R. Martinek *et al.*, "A low-cost system for seismocardiography-based cardiac triggering: A practical solution for cardiovascular magnetic resonance imaging at 3 Tesla," *IEEE Access*, vol. 7, pp. 118608–118629, 2019.

[44] J. Yao, S. Tridandapani, W. F. Auffermann, C. A. Wick, and P. T. Bhatti, "An adaptive seismocardiography (SCG)-ECG multimodal framework for cardiac gating using artificial neural networks," *IEEE J. Translational Eng. Health Med.*, vol. 6, pp. 1–11, 2018.

[45] M. Jafari Tadi, T. Koivisto, M. Pänkäälä, and A. Paasio, "Accelerometer-based method for extracting respiratory and cardiac gating information for dual gating during nuclear medicine imaging," *Int. J. Biomed. Imag.*, vol. 2014, pp. 1–11, Oct. 2014.

[46] J. M. Rubin, J. B. Fowlkes, M. R. Prince, R. T. Rhee, and T. L. Chenevert, "Doppler US gating of cardiac MR imaging," *Academic Radiol.*, vol. 7, no. 12, pp. 1116–1122, Dec. 2000.

[47] F. Kording *et al.*, "Doppler ultrasound compared with electrocardiogram and pulse oximetry cardiac triggering: A pilot study," *Magn. Reson. Med.*, vol. 74, no. 5, pp. 1257–1265, Oct. 2014.

[48] F. Kording *et al.*, "Doppler ultrasound triggering for cardiovascular MRI at 3 T in a healthy volunteer study," *Magn. Reson. Med. Sci.*, vol. 16, no. 2, pp. 98–108, Apr. 2017.

[49] F. Kording *et al.*, "Doppler ultrasound triggering for cardiac MRI at 7 T," *Magn. Reson. Med.*, vol. 80, no. 1, pp. 239–247, Dec. 2017.

[50] G. M. Nijm, A. V. Sahakian, S. Swiryn, J. C. Carr, J. J. Sheehan, and A. C. Larson, "Comparison of self-gated cine MRI retrospective cardiac synchronization algorithms," *J. Mag. Reson. Imag.*, vol. 28, no. 3, pp. 767–772, Sep. 2008.

[51] T. A. Spraggins, "Wireless retrospective gating: Application to cine cardiac imaging," *Magn. Reson. Imag.*, vol. 8, no. 6, pp. 675–681, Jan. 1990.

[52] M. E. Crowe *et al.*, "Automated rectilinear self-gated cardiac cine imaging," *Magn. Reson. Med.*, vol. 52, no. 4, pp. 782–788, 2004.

[53] R. D. White, C. B. Paschal, M. E. Clampitt, T. A. Spraggins, and G. W. Lenz, "Electrocardiograph-independent, 'wireless' cardiovascular cine MR imaging," *J. Magn. Reson. Imag.*, vol. 1, no. 3, pp. 347–355, May 1991.

[54] R. Martinek *et al.*, "A comparison between novel FPGA-based pad monitoring system using ballistocardiography and the conventional systems for synchronization and gating of CMRI at 3 Tesla: A pilot study," *IEEE Access*, vol. 8, pp. 4149–4170, 2020.

[55] M. Krej, P. Baran, and Ł. Dziuda, "Detection of respiratory rate using a classifier of waves in the signal from a FBG-based vital signs sensor," *Comput. Methods Programs Biomed.*, vol. 177, pp. 31–38, Aug. 2019.

[56] M. Fajkus, J. Nedoma, R. Martinek, V. Vasinek, H. Nazeran, and P. Siska, "A non-invasive multichannel hybrid fiber-optic sensor system for vital sign monitoring," *Sensors*, vol. 17, no. 1, p. 111, Jan. 2017.

[57] M. Krej, Ł. Dziuda, and F. W. Skibniewski, "A method of detecting heartbeat locations in the ballistocardiographic signal from the fiber-optic vital signs sensor," *IEEE J. Biomed. Health Inform.*, vol. 19, no. 4, pp. 1443–1450, Jul. 2015.

[58] Ł. Dziuda and F. W. Skibniewski, "A new approach to ballistocardiographic measurements using fibre Bragg grating-based sensors," *Biocybern. Biomed. Eng.*, vol. 34, no. 2, pp. 101–116, 2014.

[59] D. Lau *et al.*, "Intensity-modulated microbend fiber optic sensor for respiratory monitoring and gating during MRI," *IEEE Trans. Biomed. Eng.*, vol. 60, no. 9, pp. 2655–2662, Sep. 2013.

[60] M. A. Zawawi, S. O'Keffe, and E. Lewis, "Intensity-modulated fiber optic sensor for health monitoring applications: A comparative review," *Sensor Rev.*, vol. 33, no. 1, pp. 57–67, Jan. 2013.

[61] L. Mohanty and K. S. C. Kuang, "A breathing rate sensor with plastic optical fiber," *Appl. Phys. Lett.*, vol. 97, no. 7, Aug. 2010, Art. no. 073703.

[62] S. Muto, H. Sato, and T. Hosaka, "Optical humidity sensor using fluorescent plastic fiber and its application to breathing-condition monitor," *Jpn. J. Appl. Phys.*, vol. 33, no. 10, pp. 6060–6064, Oct. 1994.

[63] W.-J. Yoo *et al.*, "Development of respiration sensors using plastic optical fiber for respiratory monitoring inside MRI system," *J. Opt. Soc. Korea*, vol. 14, no. 3, pp. 235–239, Sep. 2010. [Online]. Available: http://www.osapublishing.org/josk/abstract.cfm?URI=josk-14-3-235

[64] Y.-J. Rao, D. J. Webb, D. A. Jackson, L. Zhang, and I. Bennion, "In-fiber Bragg-grating temperature sensor system for medical applications," *J. Lightw. Technol.*, vol. 15, no. 5, pp. 779–785, May 1997.

[65] J. Chen, B. Liu, and H. Zhang, "Review of fiber Bragg grating sensor technology," *Frontiers Optoelectron. China*, vol. 4, no. 2, pp. 204–212, Jun. 2011.

[66] M. Fajkus, P. Kovar, J. Skapa, J. Nedoma, R. Martinek, and V. Vasinek, "Design of fiber Bragg grating sensor networks," *IEEE Trans. Instrum. Meas.*, vol. 71, pp. 1–11, 2022.

[67] J. Nedoma, "Magnetic resonance imaging compatible non-invasive fibre-optic sensors based on the Bragg gratings and interferometers in the application of monitoring heart and respiration rate of the human body: A comparative study," *Sensors*, vol. 18, no. 11, p. 3713, Oct. 2018.

[68] Ł. Dziuda, F. W. Skibniewski, M. Krej, and P. M. Baran, "Fiber Bragg grating-based sensor for monitoring respiration and heart activity during magnetic resonance imaging examinations," *J. Biomed. Opt.*, vol. 18, no. 5, May 2013, Art. no. 057006.

[69] J. Brablik *et al.*, "A comparison of alternative approaches to MR cardiac triggering: A pilot study at 3 Tesla," *IEEE J. Biomed. Health Informat.*, early access, Jan. 27, 2022, doi: 10.1109/JBHI.2022.3146707.

[70] J. Nedoma *et al.*, "A novel FBG-based triggering system for cardiac MR imaging at 3 Tesla: A pilot pre-clinical study," *IEEE Access*, vol. 8, pp. 181205–181223, 2020.

[71] C. Larsson, L. Davidsson, P. Lundin, G. Gustafsson, and M. Vegfors, "Respiratory monitoring during MR imaging," *Acta Radiologica*, vol. 40, no. 1, pp. 33–36, Jan. 1999.

[72] M. Vegfors, L.-G. Lindberg, H. Pettersson, and P. Å. Öberg, "Presentation and evaluation of a new optical sensor for respiratory rate monitoring," *Int. J. Clin. Monit. Comput.*, vol. 11, no. 3, pp. 151–156, Aug. 1994.

[73] C. Larsson and P. Staun, "Evaluation of a new fibre-optical monitor for respiratory rate monitoring," *J. Clin. Monitor. Comput.*, vol. 15, no. 5, pp. 295–298, 1999.

[74] W. J. Yoo *et al.*, "Development of optical fiber-based respiration sensor for noninvasive respiratory monitoring," *Opt. Rev.*, vol. 18, no. 1, pp. 132–138, Jan. 2011.

[75] A. Mittal, A. K. Moorthy, and A. C. Bovik, "No-reference image quality assessment in the spatial domain," *IEEE Trans. Image Process.*, vol. 21, no. 12, pp. 4695–4708, Dec. 2012.

[76] J. Witt *et al.*, "Smart medical textiles with embedded optical fibre sensors for continuous monitoring of respiratory movements during MRI," *Proc. SPIE*, vol. 7653, pp. 516–519 Sep. 2010.

[77] J. Nedoma, M. Fajkus, R. Martinek, and H. Nazeran, "Vital sign monitoring and cardiac triggering at 1.5 Tesla: A practical solution by an MR-ballistocardiography fiber-optic sensor," *Sensors*, vol. 19, no. 3, p. 470, 2019.
[78] Z. Chen, D. Lau, J. T. Teo, S. H. Ng, X. Yang, and P. L. Kei, "Simultaneous measurement of breathing rate and heart rate using a microbend multimode fiber optic sensor," *J. Biomed. Opt.*, vol. 19, no. 5, May 2014, Art. no. 057001.
[79] J. De Jonckheere *et al.*, "Optical fibre sensors embedded into technical textile for a continuous monitoring of patients under magnetic resonance imaging," in *Proc. 30th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, Aug. 2008, pp. 5266–5269.
[80] F. Narbonneau *et al.*, "OFSETH: Optical technologies embedded in smart medical textile for continuous monitoring of respiratory motions under magnetic resonance imaging," *Proc. SPIE*, vol. 7715, Apr. 2010, Art. no. 77151D.
[81] J. Witt *et al.*, "Medical textiles with embedded fiber optic sensors for monitoring of respiratory movement," *IEEE Sensors J.*, vol. 12, no. 1, pp. 246–254, Jan. 2012.
[82] L. Dziuda, J. Lewandowski, F. Skibniewski, and G. Nowicki, "Fibre-optic sensor for respiration and heart rate monitoring in the MRI environment," *Proc. Eng.*, vol. 47, pp. 1291–1294, Jan. 2012.
[83] Ł. Dziuda, M. Krej, and F. W. Skibniewski, "Fiber Bragg grating strain sensor incorporated to monitor patient vital signs during MRI," *IEEE Sensors J.*, vol. 13, no. 12, pp. 4986–4991, Dec. 2013.
[84] D. Lo Presti *et al.*, "Respiratory and cardiac rates monitoring during MR examination by a sensorized smart textile," in *Proc. IEEE Int. Instrum. Meas. Technol. Conf. (I2MTC)*, May 2017, pp. 1–6.
[85] C. Massaroni *et al.*, "Design and preliminary assessment of a smart textile for respiratory monitoring based on an array of fiber Bragg gratings," in *Proc. 38th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. (EMBC)*, Aug. 2016, pp. 6054–6057.
[86] J. Nedoma *et al.*, "Validation of a novel fiber-optic sensor system for monitoring cardiorespiratory activities during MRI examinations," *Adv. Electr. Electron. Eng.*, vol. 15, no. 3, pp. 536–543, Oct. 2017.
[87] A. D. Wiens, M. Etemadi, S. Roy, L. Klein, and O. T. Inan, "Toward continuous, noninvasive assessment of ventricular function and hemodynamics: Wearable ballistocardiography," *IEEE J. Biomed. Health Informat.*, vol. 19, no. 4, pp. 1435–1442, Jul. 2015.
[88] S. D. Roes *et al.*, "Correction for heart rate variability during 3D whole heart MR coronary angiography," *J. Magn. Reson. Imag.*, vol. 27, no. 5, pp. 1046–1053, 2008.
[89] G. Tse, "Mechanisms of cardiac arrhythmias," *J. Arrhythmia*, vol. 32, no. 2, pp. 75–81, Apr. 2016.
[90] E. Pinheiro, O. Postolache, and P. Girão, "Theory and developments in an unobtrusive cardiovascular system representation: Ballistocardiography," *Open Biomed. Eng. J.*, vol. 4, no. 1, pp. 201–216, Oct. 2010.
[91] O. T. Inan *et al.*, "Ballistocardiography and seismocardiography: A review of recent advances," *IEEE J. Biomed. Health Informat.*, vol. 19, no. 4, pp. 1414–1427, Jul. 2015.
[92] I. Sadek, J. Biswas, and B. Abdulrazak, "Ballistocardiogram signal processing: A review," *Health Inf. Sci. Syst.*, vol. 7, no. 1, pp. 1–23, May 2019.

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