Abstract—The following paper investigates four novel methods for noncontact measurement of heart rate (HR) and consequently its derivate HR variability, an important marker of autonomic activity proven to be predictive of likelihood of future health related events. Feasibility study of basic principles is focused on measurements of signal-to-noise ratio with respect to the distance between the subject and HR sensor/apparatus. The discussed methods are divided into the following two groups: the methods measuring electromagnetic energy generated by the bioelectrical activity within the cardiac muscle (referred to as direct methods), and the methods measuring displacement of a part of the subject’s body caused by the periodic physical contractions of the heart (referred to as indirect methods). The first group is represented by a measuring device which detects changes in surrounding electric field, whereas the second group consists of measuring devices that use the Doppler effect phenomena (microwave radar, ultrasound radar) and audio signal acquired by a condenser microphone. All measuring devices were assembled and put to test. The results indicate that noncontact measuring of HR is possible, especially for distances of less than 50 cm meeting essential requirements for HR diagnostic purposes.

Index Terms—Doppler effect, electrocardiogram (ECG), heart rate (HR), HR variability (HRV), microphone, noncontact, radar.

I. INTRODUCTION

Numerous studies have shown a significant relationship between the autonomic nervous system (ANS) and cardiovascular mortality. Perturbations of the ANS and its imbalance were discovered to indicate impending cardiac diseases, which may lead to sudden cardiac death, one of the leading causes of cardiovascular mortality [1]–[3]. Furthermore, by exploring and understanding of the ANS we are able to study responses of an individual human being to particular emotional states, cognitive loads, physical loads, and other psychological characteristics [4]–[6]. This is possible by observing physiological parameters such as cardiovascular activity, respiration frequency, skin temperature, electro-dermal activity, or skin microcirculation which all have been found to be correlated with the activity of the observed person’s ANS [7].

One of the main cardiovascular parameters and at the same time one of the most commonly measured parameter is the heart rate (HR). Its derivate HR variability (HRV) has emerged as a simple, noninvasively determinable parameter for sympatho-vagal balance evaluation at the sinoatrial level. The parameter has so far been put to use in different clinical settings including diabetes, arterial hypertension, diabetic neuropathy, etc. [1], [3]. Next to this an ever increasing interest in cardiac activity monitoring is taking place in various fields and applications. Several psychophysiological experiments were published recently showing special interest to monitoring of the heart activity in real life conditions outside the controlled laboratory environment, such as studies of the psychophysiological burden of athletes [8], [9], air traffic controllers [7], drivers [10], rehabilitation patients [11], etc.

Amongst the most accurate and widely used HR determination methods are the electrocardiogram (ECG) and photoplethysmography (PPG). Although they provide reliable results when properly executed, they both are subjected to several limitations, the biggest ones being susceptibility to moving artifacts and the fact of being contact methods. As such their sensors have to be in direct and constant physical contact with the skin of the subject, making them unpleasant when used over a longer period of time or even inappropriate for specific group of patients (neonates, patients with burns, and skin diseases). Furthermore, the methods are time dependent due to drying of the conductive gel, have been known to cause allergies on the one hand (conductive gels) and can cause a certain amount of pain on the other (measuring clamp).

These drawbacks of classical HR determination methods have led researches worldwide to explore the possibilities for measuring process optimization with common goal of ridding it from the mentioned limitations. As a result, several promising novel methods and measuring techniques for noncontact measurements of cardiac activity have been researched in recent years with encouraging results. The possibility of such measurements not requiring a physical contact is thought to become a valuable tool in intensive care monitoring and perhaps also on other fields (e.g., home health care, security, sports, etc.).

The first noncontact measuring systems for human physiological parameters were carried out with radar systems in 1970s, based on the Doppler effect being able to detect respiration rates at first [12]–[14] and later also the HRs [15]–[16]. In the upcoming years several techniques for
noncontact physiological parameters were reported based on different technologies, such as laser-based technology [17], capacitively coupled sensors [18] which could compared with other technology measure the fluctuations in electrical fields in proximity of the subjects directly due to the heart activity, and lately also the image-based (RGB and thermal imaging) methods with some of them being able to detect several subjects at once [19]–[22].

In this paper, several views to the possibility of measuring HR in a noncontact manner are presented and carried out. Additionally, a feasibility study of four most promising noncontact techniques are performed and comparative study against a reference contact ECG method is made.

II. EXPERIMENTAL METHODS FOR MEASURING HR

HR can be calculated from heart interbeat interval (IBI) and is defined as the rate of occurrence of cardiac beats in a specific period of time, usually expressed in beats per minute. Although the occurrence of cardiac beats could be triggered by electrical pulses generated within the sinoatrial node (SA), the actual frequency of heart’s electrical and contractile activity is modulated by the ANS. This neural regulation causes a variability in the HR (also the resting HR), which should be relatively high in the normal physiological state of an individual and only erodes with age or progression of a disease [1], [2], [23].

Recently a number of novel noncontact methods for IBI determination were described in research papers in various fields, e.g., rehabilitation, sports medicine, transportation, and psychology. Their outcomes seem to be promising and indicate that the noncontact measurements of HR parameter are indeed possible [24]–[26]. There are, however, several limitations to the current noncontact measuring systems that need to be thoroughly investigated for optimal interpretation of the results. These limitations may be method dependent. The main common limitations are relatively small distances between the measuring probe and the human body, susceptibility to environmental disturbances, movement artifacts, etc.

Due to the functioning nature of the cardiovascular system on the one hand and the specific characteristic of the human body on the other hand, the IBI can in theory be measured in a noncontact manner directly or indirectly. First methods usually measure HR directly by observing a certain form of energy generated by the heart, whereas the second methods calculate HR based on measurements of parameters resulting from cardiac activity. Therefore, the methods from Table I, which shows a list of four novel methods that were investigated for the purpose of this research, can in principle be divided into two groups: (i) methods, based on observing the physical movement of a part of the subject’s body or the phenomena resulting from the physical movement, and (ii) methods, based on measuring the heart emitted electric fields.

### A. Indirect HR Measuring Based on Doppler Effect

While pumping blood through the cardiovascular system the heart undergoes volumetric changes during each cardiac cycle. These changes are then reflected on periodical movement of the subject’s chest which consequently contains information about its displacement as a result of respiration activity (from 4 to 12 mm with frequencies between 0.1 and 0.3 Hz) as well as due to beating of the heart (from 0.2 to 0.5 mm with frequencies between 1 and 2 Hz). The chest displacement can be measured by means of sensors with adequate resolution, as indicated in Fig. 1 [27], [28].

Radar devices have been used to sense physiologic movements since the early 1970s [12]–[16]. While unpractical at first (due to bulky, heavy, and expensive equipment), they could prove to become a welcomed monitoring solution in the future in situations where traditional methods would cause discomfort or would be difficult to apply. In principle, radar devices sense continuous electromagnetic waves that are a reflection of an active transmission. As such the radar is considered to be an active remote sensing system. It operates on different frequencies (microwave range: 2.4, 10, and 24 GHz, ultrasound (US) range: 40 kHz, etc.) and different output powers which are directly correlated to the sensitivity

### Table I

**Experimental Noncontact Methods for HR Measurement**

<table>
<thead>
<tr>
<th>Method</th>
<th>Measured Physical Quantity</th>
<th>Medium</th>
</tr>
</thead>
<tbody>
<tr>
<td>Capacitively coupled ECG (CCECG)</td>
<td>Change of surrounding electrical field due to body movement</td>
<td>Capacitively coupled electrodes</td>
</tr>
<tr>
<td>Microwave distance measurement</td>
<td>Displacement of the body</td>
<td>Microwave sensor</td>
</tr>
<tr>
<td>Ultrasound distance measurement</td>
<td>Displacement of the body</td>
<td>Ultrasound sensor</td>
</tr>
<tr>
<td>Audio signal (headphones)*</td>
<td>Change in microphone capacitance due to sound</td>
<td>Capacitor within the microphone</td>
</tr>
<tr>
<td>Audio signal (headphones)</td>
<td>Induced voltage due to sound</td>
<td>Coil within the permanent magnet in headphones</td>
</tr>
</tbody>
</table>

* The method using headphones is a contact method, but represents an innovative quasi-contact measuring technique.

Fig. 1. If it is assumed that the receiving wave comes from a single reflection on the chest surface and the surface movements are small compared with the wavelength, then according to the Doppler effect the phase change \( \Delta \theta(t) \) of the signal reflected from the subject’s chest is proportional to chest motion and scaled by the wavelength of the signal \( \Delta \theta(t) = 4 \pi \Delta x(t)/\lambda \), where \( \Delta x(t) \) is the chest displacement and \( \lambda \) is the wavelength of the transmitted signal.
environments must not exceed permissible level of density body as well as partial body irradiation for uncontrolled (according to the “IEEE Std C95.1, 1999 Edition” the whole frequencies, output power, distance, etc. [27]. The possible show the method’s ability to obtain precise results at different
transmitted through most objects except metal and water also make them appropriate for vital sign parameter monitoring through clothing, bedding, etc.

The idea of measuring HR and consequently HRV with a radar has been exercised in many studies [26]–[35]. The results show the method’s ability to obtain precise results at different frequencies, output power, distance, etc. [27]. The possible drawbacks of the method are: 1) EM radiation (achieving higher displacement resolution requires higher frequency and output power, which reflects in higher EM radiation); 2) signal susceptibility to interferences to signals operating at similar frequencies; and 3) the need of the subject to remain in still seated or lying position during the measurements.

1) Microwave Radar HR Measuring Device: The first instance of the measurement of the Doppler effect due to cardiac and respiratory activity was carried out with an off-the-shelf readily available quadrature radar microwave (μW) sensor (used in automatic door application), type RSM 2650 (by B + B Thermo-Technik GmbH), shown in Fig. 2(a). The sensor is composed of a transmitter and a receiver unit as indicated in Fig. 2(a). It has a local oscillator operating at a frequency of 24 GHz. The output RF power of the module is 40 mW. The antenna diagram has 80° angle in horizontal direction and 32° angle in vertical direction according to the data-sheet. The module has been equipped with low-noise base-band quadrature preamplifier and connected to the data acquisition (DAQ) system. The calculated power density of the used sensor at distance 30 cm from the body was $S = 0.46 \text{ W/m}^2$ (according to the “IEEE Std C95.1, 1999 Edition” the whole body as well as partial body irradiation for uncontrolled environments must not exceed permissible level of density $S = 100 \text{ W/m}^2$ at frequency $f = 24 \text{ GHz}$).

2) US Radar HR Measuring Device: Similarly to the microwave also US quadrature radar sensors can be used to detect the Doppler effect, as presented in Fig. 2(b). Two US transducers of the same type (400PT160 by Pro-Wave Electronics Corp.) were used, operating at 40 kHz. The signal at 40 kHz and reference signals for quadrature downconverter were obtained from crystal controlled oscillator operating at 10.24 MHz and divided to 80 and 40 kHz using standard HCMOS ICs. Impedance matching was done by transformers. The quadrature downconverter was built with HCMOS analogue switches (74HC4052 by Texas Instruments). Signals were amplified, low-pass filtered, and connected to the DAQ system. The measured sound pressure level (SPL) for the US radar used in the experiment was $\text{SPL} = 107 \text{ dB at distance of 30 cm from the body (the International Commission on Non-Ionising Radiation Protection – ICNIRP guideline limits the exposure to airborne US to SPL = 110 dB at } f = 40 \text{ kHz)}$.

3) Doppler Effect Signal Data Analysis: For further data analysis amplitude and frequency demodulation schemes were used on the in-phase ($I$) and quadrature-phase ($Q$) signals. The US and μW systems use local oscillator (24 GHz for μW and 40 kHz for US) and two frequency mixers, phase shifted by $90^\circ$ to obtain the in-phase and quadrature demodulated signal used for further data analysis. Amplitude demodulation was calculated as square root of the sum of squared $I$ and $Q$ channel

$$\text{AM} = \sqrt{(I^2 + Q^2)} \quad (1)$$

whereas, the frequency demodulation was calculated as the derivate of the phase angle of arctangent of the in-phase signal ($I$) divided by the quadrature-phase signal ($Q$) [36]

$$\text{FM} = \frac{d(a \tan I \div Q)}{dt} \quad (2)$$

Demodulated signals were further processed using the autocorrelation methods to obtain the HR information, described in more detail in the signal processing section of the article.

B. Indirect Methods for HR Measurement Based on Audio

In every cardiac cycle two distinctive sounds are generated, the first heart sound (S1) and the second heart sound (S2), which are produced by the closing of the atrioventricular valves and semilunar valves, respectively. By listening to and detecting these sounds with a method known as cardiac auscultation the HR can be determined [37]. One of the possibilities for noncontact HR measurement that has to the best of our knowledge not yet been described in any paper is by recording these two distinctive sounds using a condenser microphone.

1) Condenser Microphone HR Measuring Device: For the purpose of this feasibility study, a studio condenser microphone was used (RODE NT2000 by Rode). The microphone’s polar pattern was configured to bidirectional or figure 8 without a high-pass filter. This indicates that the microphone received the sound equally from both the front and the back of the element, while rejecting the sound received from the sides. The sound level that creates the same output voltage as the microphone does in the absence of sound (also known as the equivalent noise level) was 7 dBA SPL (datasheet NT2000). Theoretically, to focus merely on the sounds coming from one direction (in this case from the direction of the chest), while rejecting other sounds as much as possible, a microphone supporting the “shotgun” polar pattern would seem like a better solution. We have in fact tested such a microphone (ECM-673 by Sony), however the results did not offer any improvements compared with the NT-2000.

![Fig. 2. (a) Prototype microwave radar sensor for noncontact HR measurement. (b) Prototype US radar sensor for noncontact HR measurement.](image-url)
C. Direct Methods for HR Measurement

1) Capacitively Coupled Electrodes: Due to its typical amplitude (0.1–5 mV) and bandwidth (0.5–100 Hz) the ECG is relatively easy to measure compared with other biopotentials with conventional measuring methods, which rely on galvanic contact of the electronic sensor with the skin [38], [40]. To eliminate the need for the direct skin contact a lot of effort had been put into finding an alternative solution.

Capacitive type electrodes (CCECG) are able to detect biopotentials with an explicit gap between the sensor and the body, even through hair and clothing and can be implemented into items of daily life [38]–[45], [49], [50]. Compared with standard conductive type electrodes, the surface of these electrodes is electrically insulated and thus remains stable even in long-term applications. The sensor’s metal electrode and the body surface are capacitively coupled, forming a capacitance C

\[ C = \varepsilon_0 \varepsilon_r \frac{A}{d} \]  

(3)

where \( A \) is the effective area of the electrodes, \( d \) is the distance between the electrodes and the body, and \( \varepsilon_0 \) and \( \varepsilon_r \) the dielectric constants. As such, the capacitive type electrodes rely on detecting the so called displacement current \( I_D \), which is proportional to the rate of change of the electric field \( V \), associated with the ECG signal [38], [40].

The capacitance \( C \) depends on several factors, but usually corresponds to relatively small values from 0.1 to 10 pF. For the low-frequency measurements as is the ECG, such weak coupling requires high-input impedance of the sensor as finite input resistance would attenuate \( V \) [38], [40]. Very high-impedance nodes are very susceptible to any electro-magnetic interference from the environment and motion induced artifacts which is why the electrodes need to be actively shielded to suppress the interference.

Prance et al. presented an electric potential sensor (EPS) as a high-input impedance measurement (10\(^{15} \Omega\)) and low-input capacitance (10 fF) system with an operating bandwidth from <1 to >100 MHz, with which he was able to measure heart parameters without physical contact at a distance of up to 1 m from a clothed body in an open unshielded environment [39], [40], [46], [47].

2) CCECG Measuring Device: The electronics of CCECG was custom made and was based on the information published by several researchers (i.e., Prance, Mahdi, and Ottenbacher). The custom made sensor presented in Fig. 3 was built using the INA116 ultra low-input bias current instrumentation amplifier with bootstrapping circuit and positive feedback. The measurements were carried out with two identical electrodes positioned side by side to form a differential measuring system. Both electrodes were made out of copper clad laminate boards with a surface area of 48 cm\(^2\) (8 cm \( \times \) 6 cm) each, and overall thickness \( e = 1.5 \) mm. The HR analysis was carried out over the result of subtraction of signals from both measuring electrodes. Since we were dealing with signals of very small amplitude the CCECG needed to be protected from electromagnetic interferences. Therefore, the two copper shields were connected to the guard potential, one on the backside and one between the sensors. The reduction of parasitic effects on the electrodes was achieved by shielding the sensors. Furthermore, the electronic circuits of the sensors were isolated with a Faraday cage to minimize environment impact.

III. EXPERIMENT

A. Experimental Settings

Experiments using four noncontact HR measuring methods were carried out simultaneously at a several distances ranging from 5 up to 60 cm between the sensors and the chest of the subject. The HR values derived from the noncontact methods were compared with the values derived from the reference simplified Lead I ECG with dry electrodes. Furthermore, the results obtained in a noncontact manner were compared with each other and evaluated with a cost function. The chosen cost function was quality of acquired signal with respect to distance between the sensor and the human body. Quality of signal was determined by calculating signal-to-noise ratio (SNR).

The experiments were conducted on one of the authors wearing a single layer clothes (T-shirt). The subject was seated on a standard office chair with plastic frame and cotton seat and was asked to minimize the movement while breathing normally for the duration of the experiment. The duration of a measurement at a particular distance was 70 s. To also investigate the susceptibility of devices to extraneous electromagnetic fields, no special care was taken to prevent usual power-line interferences of electric and as well magnetic nature (power line, electric devices, etc.). The HR of the subject was measured with a total of six different measuring methods simultaneously: 1) ECG (reference contact method); 2) CCECG; 3) microwave radar; 4) US radar; 5) audio signal microphone; and 6) headphone audio contact method (Fig. 4).

The output signals from all measuring devices were sampled by means of a high-performance DAQ system UEILogger 600 and DNA AI-217 AD board. The recordings of all measuring systems were sampled simultaneously with 24-bit AD conversion at sampling frequency of 2 kHz.

B. Reference Lead I ECG

The discussed novel measuring methods within this paper were compared with a simplified contact Lead I ECG
Fig. 4. Experiment was conducted in an unshielded environment in close proximity to several sources of line noise. The four noncontact experimental methods were positioned at the same distance from the subject for each measuring cycle. The signal was simultaneously recorded with the reference Lead I ECG and a headphone audio contact method.

Fig. 5. Measuring setting for reference method evaluation process. The signal from the Lead I ECG recording device was the voltage between the left arm and the right arm (I = LA-RA). The signal from the 3-lead commercially available ECG recording device was the combination of voltage between left arm, right arm and left leg (I = LA-RA; II = LL−RA; and III = LL−LA).

recording device which served as a reference method. To determine its accuracy, an evaluation study by means of direct comparison against a commercially available Lead III ECG monitor ECG1000C (by Biopac Systems, Inc.) was carried out. The signals from both were recorded from electrodes positioned on both wrists and the electrode on the left leg, as indicated in Fig. 5. During the process the subject was seated still with hands resting on his thighs.

The Lead I ECG recording was carried out using two off-the-shelf antistatic bracelets. The elastic bracelets were used to ensure firm contact of the electrodes with the skin, thus enabling a reliable attachment of the sensors and minimizing the possibility of noise due to movement artefact during the measuring interval. Since no conductive gel was applied the two electrodes can be considered as dry contact. The recorded signal was amplified with INA118 instrumentation amplifier (by National semiconductor) with input impedance $Z_{in} = 10 \, \text{M}\Omega$ and acquired by the DAQ system. Although the DAQ system provides input impedance greater than $100 \, \text{M}\Omega$ the amplifier was used to reduce the crosstalk to/from other sensors.

For the purpose of evaluation process a 30 s recording of the cardiac activity was taken from both measuring devices simultaneously. Fig. 6 shows a 2.5 s segment of the recording after the band-pass (5–35 Hz) and notch (50 Hz) filter were applied to the Lead I ECG. The overlapping of both signals was high and the peak of the signal was very distinctive making it relatively easy to be detected and used for HR analysis, as can be seen in Fig. 7.

In Table II the statistics of the HR comparison is given, showing a negligible difference between the signals and thus proving the suitability of our simplified Lead I ECG device for the purpose of this paper as a reference signal.

C. Signal Processing

The information about cardiac activity can be presented in different spectrum ranges and depends on the measuring method used. For optimal performance and signal seeking options different filtering needs to be applied for different measuring methods. Based on conducted experiments...
the settings shown in Table III were recognized as the optimal.

After demodulation and filtering in the case of microwave and US radar and custom signal filtering to remove noise from the biomedical signals, local peaks of each waveform were searched for using an autocorrelation method. The autocorrelation method was performed using existing function blocks within dedicated PC software, in our case MATLAB and Labview. The local peaks were thought to correspond to the R peaks acquired by the reference ECG method.

Our measurements however show a certain time delay between the two concepts, as can be seen in Fig. 8. This is the result of multipolar nature of the cardiac electric field dynamics as well as the pulse transit time, leading to temporal and spatial dependence of the measured signal. Additional minute time delay is also contributed due to the use of different filters (see Table III for filter settings). The IBI however remains unchanged when measuring the biomedical parameter with the specific measuring method at the same location.

Generally speaking, in a noncontact device the measured signal which is caused by cardiac activity must be stronger than other noise sources picked up by the sensor to make reliable and accurate measurements. To evaluate the accuracy of the novel noncontact methods in relation to the distance between the subject and the measuring system, a SNR cost function was used. In our experiments the SNR was calculated as a ratio between signal (peak-to-peak amplitude of the acquired signal when the physiological signal was detected) and noise (peak-to-peak amplitude of the signal acquired signal when there was no physiological signal detected, in other words signal between two heart beats), as indicated in Fig. 9(a).

IV. RESULTS

Fig. 10 represents a 10 s segment obtained with one of the novel methods (US, amplitude demodulation). Within this measurement (as well as within other novel method measurements) two physiological parameters can be seen: 1) the respiration at lower frequencies (indicated with a dashed line in Fig. 10) and 2) the HR at higher frequencies. These two signals could easily be separated by applying
additional filters, an activity not included in our feasibility study experiments.

As expected the novel noncontact methods are greatly affected by several factors. Since all of the recordings were conducted simultaneously and within a relatively short time span (~1 h, taking into consideration the 70 s measuring time interval per setting, several settings measuring at different distances and the necessary preparation needed for each setting) the deviation of environmental variables, such as relative humidity, temperature, ventilation, etc., were considered to be kept at a minimum level. If assumed that the subject could indeed maintain the same level of psychological calmness during the whole time, we can conclude that the only considerable influence on the results obtained in different settings was the distance itself between the sensors and the subject. Three sets of measurements carried out on different distances (5, 15, and 40 cm) between the noncontact measuring sensors and the subject are presented in Figs. 11–13.

Fig. 11 shows the waveforms of signals acquired at the distance of 5 cm between the electrodes and the subject. The waveform marked with “ecg” in the top left part of the figure represents the electrical signal obtained with the reference Lead I ECG contact recording device after rejecting the line frequency noise and band pass filtering. The signal form is of close resemblance to the standard ECG (see Fig. 6) with clearly recognizable R peaks over the rest of the signal. The second waveform marked with “mic” represents the unamplified and filtered audio signal obtained with a condenser microphone (with figure 8) bidirectional pattern. The signal shape is a consequence of heart-beat audio signal and surrounding audio noise. The two most distinctive heart sounds often described as a “lub” (S1) and a “dub” (S2), which are caused by the closing of the AV valves and semilunar valves, respectively, coincide with the R and the T wave on the ECG recording appearing as an oscillation of the signal on the presented waveform. The third waveform marked with “hp” was carried out with dynamic headphones (AKG141 MKII). Signal shape is a consequence of heart-beat audio signal picked up with headphones functioning as a microphone, head movements...
and surrounding audio noise. The next waveform, marked with “cc eccg” represents the signal detected from fluctuations of the EM field in the vicinity of the subject’s chest. The signal peaks correspond to the heart beats delayed and oscillated due to mechanical movements of the heart and torso. The amplitude demodulated signal of the US device marked with “us a” and microwave marked with “μW a” show the amplitude of the reflected signal from the subject’s chest. US and microwave frequency demodulated signal (“us f” and “μW f”) represent the frequency variation of the received signal from the subject’s chest due to the Doppler shift. Compared with the amplitude demodulation signal the frequency demodulation signal holds also the respiratory information superimposed to the HR information, which can be detected on the waveforms as amplitude fluctuation of the signal (see also Fig. 10 for clearer representation).

Figs. 12 and 13 show the waveforms acquired by the noncontact measuring sensors from distance of 15 to 40 cm to the subject’s chest, respectively. Apart from the two contact methods (“ecg” and “hp”) the amplitude of all signals is decreasing along with the distance between the sensors and the subject which is to be expected. At larger distances some measurements contain more noise than useful signal, making the relevant information extraction hard, unreliable, or even impossible. The two methods that appear most dependent to the explicit air gap between the measuring sensor and the subject are the condenser microphone method and the CCECG method (Fig. 14).

As a result of our study a correlation of SNR for every method with distance is determined and shown in Fig. 14. From the results we can conclude that SNR characteristics of different devices depend on the method itself, its characteristics and other factors, such as type of modulation (amplitude, frequency) of signal conditioning, etc. For distances below 10 cm the method with the most convenient SNR value is microwave radar method, while above 10 cm distances US radar methods shows the best results.

V. DISCUSSION

Four novel noncontact methods for determining the human HR were investigated, out of which one was direct and three were indirect. The direct method for measuring the HR was monitoring the fluctuations within the electric field generated by the heart activity. Since the searched for signals are of very small amplitude an additional high-voltage generator (1 kV) was applied (via insulated wires) near both sensors to achieve a higher sensitivity. The acquired information with the CCECG method proved decent at small distances. However, at measuring distances of over 10 cm the ambient noise became dominant, affecting the signal SNR and as such making the useful signal extraction difficult or impossible. Three indirect methods measured chest displacement by means of different techniques, i.e., microwave and US radar technique and by means of an audio technique using a condenser studio microphone which was according to our knowledge not discussed in any paper so far. While being a new measuring method, we can conclude that it could be useful at relatively short distances (approximately up to 10 cm). Similarly to the CCECG method, however, the environmental noise rapidly became dominant at larger distances, making the obtained waveform indistinguishable. The US and microwave radar techniques have proven to be most robust amongst the tested methods and capable of extracting the HR information at the largest distances. Next to the cardiac information also the respiration information can be detected in frequency demodulated signals of both methods.

All of the discussed noncontact measuring methods have a certain time delay of the local peaks which correspond to the R wave in the reference Lead I ECG signal. This time delay is mainly thought to be of mechanical origin.

Some advantages and some limitations of the methods were discussed in this paper. For example, in contrast to other novel noncontact methods the microwave and US radar systems were more capable of detecting motion artefact due to cardiac and respiratory activity from larger distances with signal still being suitable for HR evaluation. Furthermore, the sensor could be camouflaged making it possible to perform a measurement on a human being without his awareness of the procedure, thus avoiding potential psychophysiological effects. The measurements were otherwise less affected by electromagnetic interferences from the environment as compared with other methods. The sensor was however prone to movement artefacts of the subject. Furthermore, both measuring methods are considered as active and thus include US and RF radiation, respectively.

The measuring method with a microphone is a novel method previously not yet discussed in any paper. The results obtained with the method at short distances yield outcomes at short distances. The limitation of the microphone method was that it could be effectively used at relatively short distances (<10 cm) and the measurements were affected by several factors, such as drafts, air circulation, and movement of the subject relative to the sensor.
Another measuring method which was tested in the course of the experiment was the method with headphones. While this is still a contact method which offers a good signal we believe it could be found convenient when integrated into a pleasant everyday experience, such as listening to music.

The last of the discussed noncontact measuring methods (CCECG) is a passive method with no need for electromagnetic radiation. While it provided good results at relatively short distances (up to 10 cm) the results obtained at longer distances contained more noise than useful signal. Furthermore, to achieve higher sensitivity of the sensors a local electrical field was applied in the close proximity of the measuring electrodes.

Our study was feasibility oriented and we were not interested in accuracy, precision or other metrological properties of the new devices. Our future work will therefore focus more on signal processing, accuracy validation and robustness of the measurements. The bio-signal processing is a challenging task. There are however many algorithms already developed for automated processing and relevant information extraction, which can be done in time or frequency domain with different methods, such as detrended fluctuation analysis (DFA) method [51] and spectrum processing based on a Fourier transform [52]. Future studies will include metrological evaluation of the measuring results for every method. Measuring errors and measuring uncertainty of different measurement will be estimated to ensure proper traceability to higher metrological standards. Uncertainty budget calculation and traceability will be established by calibration by comparison against a calibrated 12-lead reference ECG device.

VI. CONCLUSION

Our feasibility study showed that all four methods were capable of reliable measurement of IBIs and were suitable for HR determination over distances from 5 to 40 cm (method dependent) in common laboratory conditions, with no special electro-magnetic shielding.

The novelty of the proposed direct method is the artificially generated electric field in the sensor proximity which enhances the signal obtained with CCECG sensor. The novelty of indirect methods on the other hand is the possibility of parallel implementation of all three methods (US, microwave, and condenser microphone), thus achieving a more robust and reliable measurement.

For future work it is recommended that after basic feasibility study an analysis of metrological parameters, such as accuracy, certainty and sensitivity, will be thoroughly investigated, followed by a clinical evaluation of the measuring methods. The emphasis of HR measurement will primarily be focused on HRV measurement.

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REFERENCES


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