A Wireless Joint Communication and Localization EMG-Sensing Concept for Movement Disorder Assessment

Stefan Brückner, Jasmin Kolpak, Fabian Michler, *Member, IEEE*, Nikita Shanin, *Graduate Student Member, IEEE*, Robert Schober, *Fellow, IEEE*, Amelie Hagelauer, *Senior Member, IEEE*, Robert Weigel, *Life Fellow, IEEE*, Heiko Gaßner, Jürgen Winkler, Björn M. Eskofier, *Senior Member, IEEE*, and Martin Vossiek, *Fellow, IEEE*

***Abstract*—Real-time sensory recording of the musculoskeletal system function is an important tool for the diagnosis, treatment planning, and optimal treatment execution of diseases, such as Parkinson's disease and osteoarthritis. This article presents a new wireless joint communication and localization electromyography (EMG)-sensing concept. An on-body sensor beacon measures EMG signals and wirelessly transmits them. At the same time, the spatial position and movement of the beacon is determined with high precision in real time using these transmitted radio signals. The seamless integration of multiple sensors avoids the need to synchronize and individually set up multiple independently operating sensors. An outstanding feature of the radio localization approach is that it does not require proprietary ultra-wideband signals or complicated time synchronization protocols, allowing for small and energy-efficient implementation. To demonstrate this novel concept, a wireless 3D-localizable EMG sensor was developed and experimentally evaluated. This new type of sensing concept allows, for the first time, the time-synchronous measurement of muscle activity and the underlying movement of the associated body part.**

***Index Terms*—Array signal processing, EMG, localization, radar tracking, radio communication.**

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Stefan Brückner, Jasmin Kolpak, Fabian Michler, Nikita Shanin, Robert Schober, Robert Weigel, and Martin Vossiek are with the Department Electrical Engineering, Friedrich-Alexander-University Erlangen-Nürnberg, 91058 Erlangen, Germany (e-mail: stefan.brueckner@fau.de; jasmin.kolpak@fau.de; fabian.michler@fau.de; nikita.shanin@fau.de; robert.schober@fau.de; robert.weigel@fau.de; martin.vossiek@fau.de).

Amelie Hagelauer is with the Fraunhofer Institute for Electronic Microsystems and Solid State Technologies EMFT, 80686 München, Germany (e-mail: amelie.hagelauer@emft.fraunhofer.de).

Heiko Gaßner and Jürgen Winkler are with the Department Molecular Neurology, University Hospital Erlangen, 91054 Erlangen, Germany (e-mail: heiko.gassner@uk-erlangen.de; juergen.winkler@uk-erlangen.de).

Björn M. Eskofier is with the Department Artificial Intelligence in Biomedical Engineering, Friedrich-Alexander-University Erlangen-Nürnberg, 91052 Erlangen, Germany (e-mail: bjoern.eskofier@fau.de).

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I. Introduction

MOVEMENT disorders such as Parkinson's disease and osteoarthritis are some of the most common age-related diseases worldwide. These diseases are associated with significant disability and morbidity [1], [2]. For an effective treatment of these disorders, early and accurate diagnosis is crucial. According to [3], classical 3D-gait analysis is the state-of-the-art in clinical environments to determine the pathological degree of movement restriction, monitor the related progression over time, and assess the impact of prevention and rehabilitation measures. Thereby, apart from visual inspection, suitable sensor concepts are required to provide additional sensory data and quantitative measures. To analyze gait patterns, motion capture systems are necessary. Diagnostic systems currently in use are based on inertial measurement units (IMUs) [4], [5], [6], [7] or optical systems [8], [9], [10], [11]. The combination of classical 3D-gait analysis with additional information about the muscle activity and their temporal behavior collected by a synchronized EMG sensor enables the determination of muscle activation or fatigue information in specific gait phases. This leads to a novel powerful tool that is able to obtain patient-specific gait pattern models with unique information about the causality of gait alteration [12]. From a clinical viewpoint, this enables physicians to draw a comprehensive picture of symptoms for diagnostic purposes and clinical decision-making by using them over a longer period of time in real-life scenarios. In neurodegenerative disorders such as Parkinson's disease, motion tracking and EMG devices are considered to be used for monitoring therapy response and motor fluctuations as well as for improving rehabilitation interventions [13]. In particular, it is known that the combination of kinematic and EMG measures provides detailed clinically meaningful outcomes for gait analysis in neurological populations [14]. To reduce the application complexity and setup times in everyday medical practice, the corresponding sensor systems should enable continuous monitoring, be easy to use, and be able to monitor common movements in a natural, non-disturbing real-world setting. The sensors should also be small, lightweight, and connected wirelessly while being powered by a battery or via energy harvesting to ensure complete freedom of movement.

A motion-capturing setup that combines IMU data with data recorded by a Bluetooth EMG sensor was investigated in [15]. While being positively validated, the problem of drift concerning the IMU remains a huge problem [16], [17]. As shown in [4], skeleton models help to partially solve these problems, but multiple body parameters are necessary for posture estimation. Incorrect or inaccurate parameters strongly influences the position estimation, making IMU systems questionable for precise medical analysis [18], [19]. More robust results are achieved by combining EMG with optical motion-tracking systems for absolute positioning. However, the large setup effort for the markers and sensors, the need for free skin or tight clothes, and high costs inhibit the practical usability of such systems. To overcome the above limitations, classical localization techniques using the already available radio frequency (RF) signals not only for data transmission but also for motion tracking are quite promising. However, vast effort has to be made, to avoid the interference between data transmission and the localization [20], [21], [22]. This becomes even more challenging for wireless multi-sensor systems, where the sensors share a single frequency band, where a tight timing protocol has to be implemented and sensors have to be synchronized between each other. The contribution of this publication is the introduction of a novel EMG-sensing concept, where a sensor transmits the captured data wirelessly and without restrictions on modulation while being absolutely localized in 3D using the same signal, as well as the description and validation of a proof-of-concept hardware setup. The developed concept enables an easy to set up and use, cost-effective, and reliable diagnostic tool for gait analysis which does not need a manual synchronization of the captured data.

The proposed system concept with an explanation of its benefits is presented in Section II. The hardware implementation of the localizable sensor is described in Section III, followed by the presentation of the receiver hardware and localization algorithm in Section IV. The experimental verification of this concept is presented in Section V, concluded by final remarks and outlook in Section VI.

II. Proposed System Concept

As discussed in Section I, a diagnostic system using RF localization offers several advantages and results in a better usability in terms of setup and use for the assessor as well as the patient compared to the traditional methods with an EMG and IMU or optical system combination. Thereby, a joint RF communication and localization system is preferable, where an on-body hardware, here referred to as beacon, measures the EMG signals and, at simultaneously operates as a node that is localized by multiple base stations for gait or body movement analysis. This approach must overcome various specific challenges, such as tracking 12 key points, three for each extremity of the patient, simultaneously performing gait analysis and, at the same time, recording and processing multiple EMG signals. The general key requirements of the overall system derived from these challenges are as follows:

1) **Synchronization of EMG data with position details:** As stated, the combination of motion tracking and EMG provides great benefits for diagnosis and therapy response. However a precise temporal linking between muscle stimulation (EMG) and muscle movement (motion tracking) is required for a meaningful gait analysis [14]. As an example, [23] evaluates the EMG signals directly before a freezing caused by Parkinson's disease occurs. Also, the muscle activation timing in relation to the gait cycle in [24] requires precise synchronization between the measurement devices, which either has to be done manually or desirably automated.
2) **Small size and low complexity of the transmitter:** The goal is to evaluate human motion and, particularly, whether there is a deficit. As a result, the beacon must not affect the motion itself, even if it is worn over a period of time in an everyday environment, and therefore, needs to be small in the first place. Considering its power supply, a wireless solution is strongly preferred to a wired one since cables severely influence and restrict natural motion sequences and impose a large overhead when attaching the system. Therefore, one key aspect regarding the size is the utilized battery. To keep the battery as small as possible, a low power consumption achieved by appropriate hardware and software design is necessary. This leads to the requirement of low computational effort in the beacons and sleep times between the EMG-sampling phases. In addition, an efficient antenna design needs to be considered. In particular, ultra-wideband (UWB) antennas are often sized multiples of wavelengths, which inhibits a compact beacon design.
3) **Synchronized position estimation for each beacon:** For assigning different gait cycles, as well as the whole body posture, multiple beacons have to be considered in regard to each other. For fast-paced movements, even small time delays impose errors in posture determination.
4) **High accuracy of the localization itself:** Just as temporal errors strongly influence the posture evaluation, a spatial accuracy of at most a few centimeters is necessary to assign the correct pose.

The most obvious approach for meeting these requirements is based on a localization with the evaluation of distance information between a beacon and multiple receivers, such as time difference of arrival (TDOA) or round trip time of flight (RTOF). However, several drawbacks make this solution far from ideal: The localization requires the beacons to transmit predefined signals that can be evaluated by the receivers to obtain distance information. Consequently, the localization signal needs to have a specific shape, e.g. frequency-modulated continuous-wave (FMCW) or UWB pulses. These specific-shaped signals are difficult for data transmission, and thus for precise localization, a second system would be required for the communication, which would then have to be synchronized with the position information. Furthermore, localization with high accuracy requires broadband signals associated with comparatively complex antenna structures. Additionally, the need to locate multiple beacons at the same time leads to interfering signals.

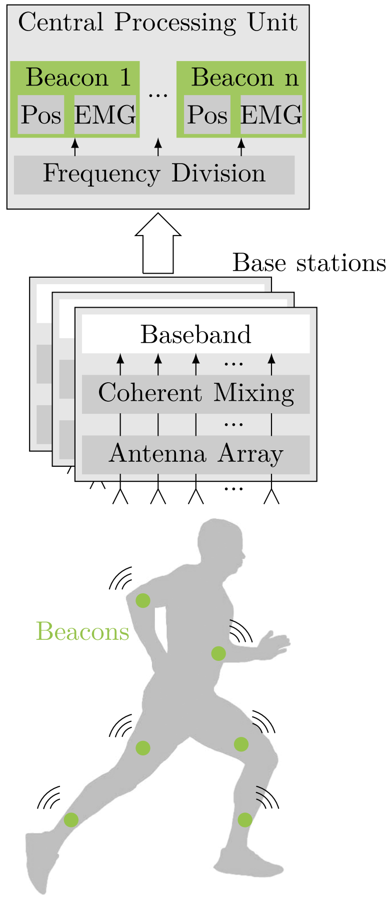

Fig. 1. Proposed sensor setup consisting of multiple beacons, attached to the human body, transmitting captured EMG data as well as multiple base stations coherently converting the RF signal to a baseband signal. A central processing unit divides the baseband signal into multiple frequency channels to separate the beacons' signals and calculate the corresponding position and EMG data for each beacon.

In turn, this requires a complex transmission protocol with time-division multiple access (TDMA) to enable information reception at each beacon which makes the system design even more complex. However, then, the position information may still not be recorded simultaneously, which compromises its suitability of such systems for gait analysis.

To meet the above requirements, the proposed concept shown in Fig. 1 uses the same RF signal for data transmission and localization without interference. Therefore, the beacons are implemented combining an EMG sensor and an RF transmitter employing conventional low-bandwidth modulation schemes on a single RF carrier to send data to multiple base stations. These base stations are distributed around the capturing area and consist of spatially distributed antennas that all coherently receive and mix the beacons' signals to an intermediate frequency (IF), which is then digitized and sent to the central processing unit. For low-bandwidth modulation, each beacon uses its own frequency band, and the signals of all beacons are transmitted simultaneously and separated at the central processing unit by bandpass filters. In contrast to typical localization approaches, here, the concept of phase difference of arrival (PDOA) is applied to the spatially distributed antennas at the base stations. Utilizing a Holographic Extended Kalman Filter (HEKF), recently introduced by the authors in [25], [26], the beacons are localized by using the same RF signal as for data transmission. To this end, the HEKF recursively evaluates the relative phase of the RF signal by an Extended Kalman Filter (EKF). As shown in [27], the resulting localization accuracy is not dependent on the bandwidth, as in conventional runtime-based localization schemes, but is determined by the size of the apertures with regard to the beacon-base station distance. This makes the HEKF a perfect candidate for precise indoor localization. By using the same signal for communication and localization, the received data is inherently synchronized with the position information, which is a great benefit over traditional measurement approaches with multiple different systems, where manual synchonization is necessary. By using narrowband signals, the separation of the different beacon signals does not have to be done in the time domain, but frequency-division multiple acess (FDMA) can be used for separation. Each beacon can continuously transmit the acquired data in real time or with a small predictable delay. Therefore, on the one hand, a direct time link between the EMG and position data of the beacon is achieved. On the other hand, this continuous transmission enables the HEKF to track all beacons simultaneously in a time-coherent manner. Finally, as the implemented beacons can transmit the captured data directly, they do not require any form of synchronization, which yields low computational effort on the beacon side. Thus, by eliminating inefficient broadband modulation, an energy-efficient beacon can be built using only a small battery and a small antenna. Overall, in contrast to traditional approaches or other RF-solutions,this concept meets the above requirements, and thus, represents a promising solution for gait analysis with combined EMG sensing.

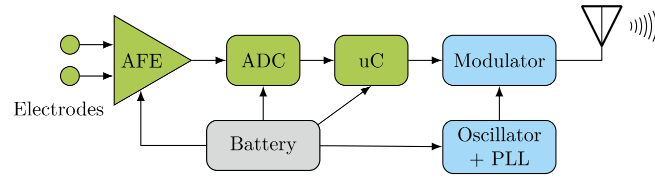

Fig. 2. Hardware overview of the beacon consisting of the low-frequency parts in green: analog front-end, which capture the EMG signal through two electrodes and amplifies it in order to be digitized in the following analog digital converter and processed by the microcontroller. The RF parts, i.e. the 24 GHz oscillator and the following modulator, which is driven by the microcontroller, are shown in blue. The power is supplied by a battery, depicted in gray.

III. Beacon Implementation

In this work, a non-miniaturized demonstrator of the beacon was constructed. Although this demonstrator does not yet meet the requirements for a sophisticated medical device due to the high power consumption and large size, it still serves as a prototype to validate the concept and its claimed benefits. The related block diagram is shown in Fig. 2. The sensory part of the beacon consists of a single-channel surface electromyography (sEMG) block with electrodes, an analog front-end (AFE), and

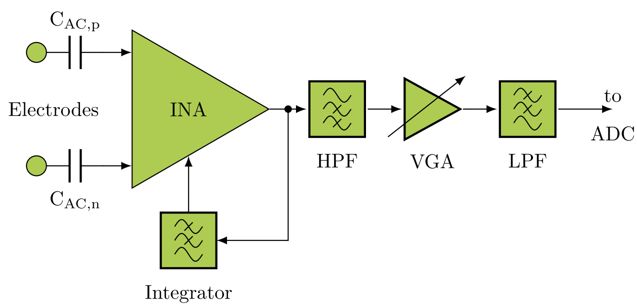

Fig. 3. Block diagram of the sEMG sensor including two AC- coupled electrodes, instrumentation amplifier, feedback path with low-pass filter, as well as a high-pass and low-pass filter and variable gain amplifier to optimize the gain for analog-digital conversion.

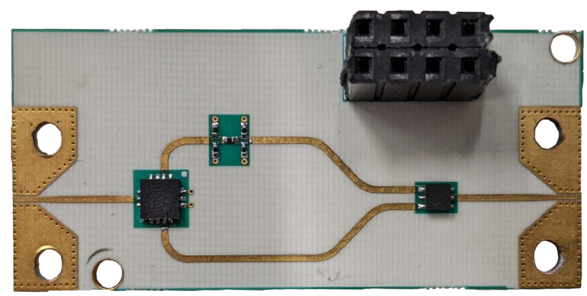

Fig. 4. 24-GHz low-cost modulator, consisting of SPDT-RF switch, alternating the signal between a microstrip line and attenuator, as well as a combiner.

an analog-digital-converter (ADC). For wireless data transmission, an RF transmitter is realized by using an oscillator with a phase-locked loop (PLL), a modulator, and an antenna. The data pre-processing and encoding as well as the modulation of the RF signal are performed by a low-power microcontroller.

While most of the EMG sensor topologies reported in the literature [28], [29], [30], [31] are based on a third electrode that serves as a reference to the human body, the sEMG sensor used in this setup requires only two electrodes for measuring the potential difference between two points on a muscle to keep the sensor as small and simple as possible for the end user.

For sEMG signals, the relevant differential voltage amplitudes detectable on the skin surface are in the range between a few µV and 5–6 mV [31]. These analog voltages must then be digitized to process information and transmit data to the base station. Therefore, an ADC is necessary, which usually samples analog signals in the order of a few volts. Hence, the recorded weak sEMG signal must first be amplified by the AFE to fit the input range of the ADC,i.e., with gain factors of 100 to 2000 [31]. Since the relevant information content of sEMG signals lies in the frequency range between 0 Hz and 500 Hz, the strong interference caused by the power line at the frequency of 50/60 Hz is the most challenging issue for the implementation of the AFE [32]. This also applies to battery-powered devices, especially in the absence of a reference electrode. The concept of the realized AFE is shown in Fig. 3. In the proposed setup, adhesive silver-/silver-chloride (Ag/AgCl) electrodes with a recording diameter of 4 mm are used to connect the beacon to the proband. From an electrical perspective, the resulting interface between the electrodes and the human skin corresponds to a complex impedance in the range of hundreds of kΩ to 1 MΩ [33], thus forming a voltage divider with the input impedance of the following amplifier. To avoid an additional attenuation of the already weak sEMG signal, the optimal amplifier input impedance according to [33], [34] is in the range of 100–1000 MΩ.

Together with the requirement of a common-mode rejection ratio (CMRR) higher than 100 dB at 50/60 Hz related to power line interference caused by parasitic coupling effects between the mains and the human being leads to the use of an instrumentation amplifier (INA) as a subsequent pre-amplification stage [34]. To suppress DC offset voltages resulting from polarization effects of the electrodes, the input signals captured by the electrodes are AC-coupled by the series capacitors $C_{AC,p/n}$ in Fig. 3. The pre-amplified signal is then filtered due to the limited EMG content of interest, as mentioned earlier. Therefore, an active second-order high-pass filter with a 3 dB cutoff frequency of 7 Hz and a passive second-order low-pass filter with a 3 dB cutoff frequency of 500 Hz are implemented. In case of internally generated noise and DC offset voltages, an analog integrator with low-pass behavior connected to the reference input of the INA is realized, which functions as an additional high-pass filter to the pre-amplified signal. Due to the wide sEMG input voltage range of 10 µV to 10 mV, depending on the muscle of interest and the anatomical constitution of the subject, a second amplification stage with variable gain is necessary: the VGA. With the proposed AFE topology, the total gain factors between 46 dB and 98 dB are possible. Once converted into the digital domain by a 12-bit ADC, the upper eight bits of this digitized data are then processed by a microcontroller, which cashes multiple EMG samples, performs error correction coding, and adds an M-sequence for synchronization with the receiver. Each single transmission transfers 30 uncompressed EMG samples. For error correction, a Reed-Solomon implementation, namely RS(36,6), is used, i.e. for every 30 data bytes, an additional six redundancy bytes are calculated. This allows for the correction of up to three bit errors and detection of up to six bit errors [35]. Additionally, the first 32 bits are reserved for an M-sequence, resulting in 320 bits for each transmission, which are sent at the data rate of 50 kbit/s. To transmit this data, a continous-wave (CW) source is modulated with a switchable attenuator. The 24-GHz CW signal is generated by an oscillator, which is controlled by a PLL referenced by a quartz crystal. For the modulator, a single pole double throw (SPDT)-RF switch is used, which is digitally controlled by the microcontroller. Depending on the switch position, the signal is guided through a −10 dB attenuator or a microstrip line. Both are combined again with a power splitter/combiner, resulting in a binary amplitude shift keying (ASK) modulation scheme. The realized modulator is shown in Fig. 4. Compared to state-of-the-art modulators using an I/Q-mixer, this design stands out through its ease of implementation and low hardware complexity, which makes it suitable for low-cost sensors. One disadvantage of this scheme is its low spectral efficiency because of the binary-ASK modulation. In Fig. 5, the signal in one transmission block is shown.

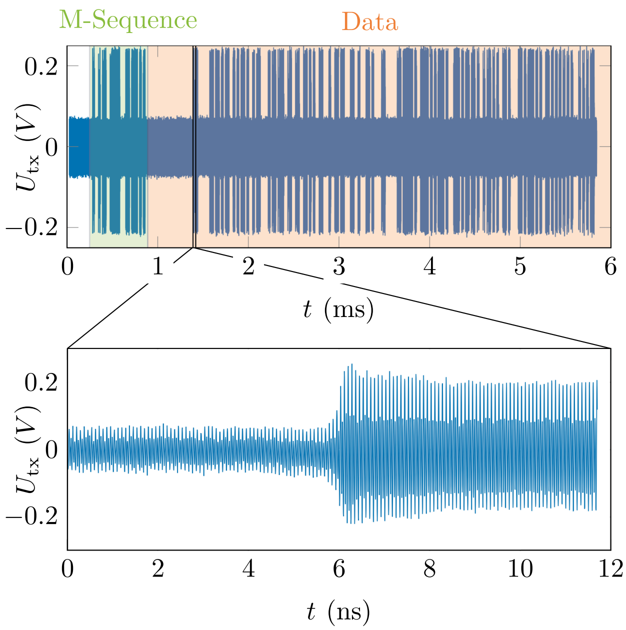

Fig. 5. View of the transmitted signal with 24 GHz carrier frequency and 2-ASK modulation. The first 32 bits are an M-sequence to find the beginning of a transmission, followed by 288 data bits, encoded in a RS(36,6) implementation. A detailed view is presented for one 0-1 transmission.

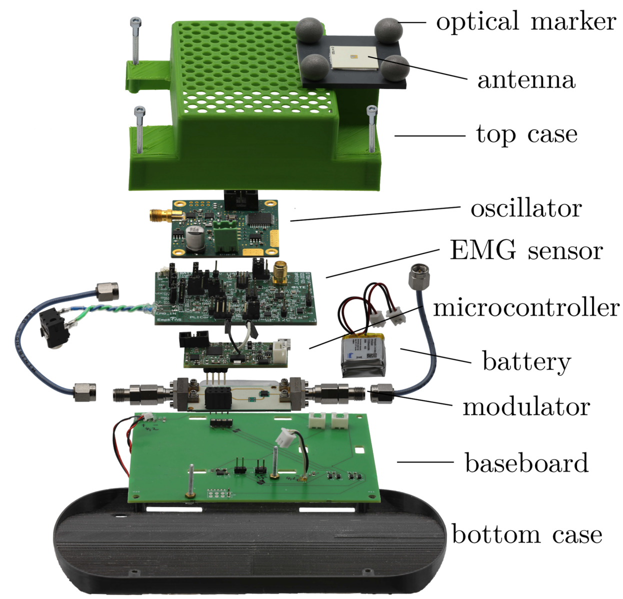

Fig. 6. Exploded view of the prototype sensor, comprising of a case with a patch antenna and four optical markers, the center point of which coincides with the antenna position, a 24-GHz oscillator, an EMG sensor, a microcontroller, a battery, and a modulator, as well as a baseboard, and a bottom case with mounting holes.

Finally, a patch antenna is used for transmitting the signal. A two-cell battery powers the AFE, ADC, microcontroller, and oscillator. The total power consumption is 1.3 W, which is not yet a concern as this is a proof-of-concept. The individual components are connected through a baseboard and housed in a 3D-printed case for mounting on a participant for evaluation. In Fig. 6, an exploded view of the realized beacon is shown.

IV. Base Stations and Wireless Localization

To receive the data and localize the beacon, multiple base stations are distributed around the capture area to cover the beacon from at least two angles. Each receiver generates its own local clock, and thus, only a coarse synchronization between the receivers and no synchronization between the beacons and the receivers is necessary. The aperture consists of multiple antennas arranged to enable unambiguous position-dependent phase information of closely spaced antennas, as well as iteratively larger antenna distances to improve accuracy while retaining unambiguity, as explained in [26]. For each antenna, the receiving signal is mixed coherently at the base station to a complex valued IF signal, followed by a low-pass filter for mirror-frequency suppression and anti aliasing. An ADC digitizes the signal, which is then sent over Ethernet to the central processing unit. The separation of different beacons is then digitally performed by using bandpass filters.

For data recovery, the envelope of the digital IF signal is calculated and a cross-correlation with an M-sequence is performed to find the precise start of a data transmission. Now, with the predefined symbol-rate, each bit can be determined. A Reed-Solomon decoder is implemented to correct bit errors in the transmission. Finally, a local timestamp is added to the data to precisely link it to the localization information.

For localization, the IF signals of one beacon at each receiving channel are cross-correlated with each other. This yields a position-dependent phase, which is then evaluated in the HEKF. Compared to a conventional angle of arrival (AOA) localization algorithm, which fist calculates the angle of the signal source and then combines multiple AOAs via triangulation to find the position, the HEKF does not rely on farfield assumptions, and therefore, can be used in the nearfield regarding the aperture size, minimizing the errors that occur through the nonlinear operation of calculating the angle from phases, as shown in [36].

In the following, a brief description of the HEKF is presented, whereas in [26] the algorithm is derived in detail. First, an assumption on the movement model and the determination of the measurement model have to be made. For the movement model, in general, any assumption can be made; here, a linear movement was chosen due to its simplicity. Thus, for a given state of position and velocity, a prediction of the next state is made through linear extrapolation. For the measurement model, it is assumed that in measurement step k an unsynchronized beacon transmits the signal

$$s_{\mathrm{B},k}(t) = A \cdot s_{\mathrm{TX,BB},k}(t)\mathrm{e}^{\mathrm{j}(\omega_{\mathrm{TX}}t+\varphi_{\mathrm{B},k})} \tag{1}$$

with amplitude A, a so far unconstrained baseband signal $s_{\mathrm{TX,BB},k}(t)$, the roughly known angular frequency ω_{TX}, and the phase offset $\varphi_{\mathrm{B},k}$, which is unknown and independent for each step. At the ith antenna of the receiver, the signal $s_{i,k}(t)$, which is the transmitted signal delayed by $\tau_{i,k}$ (as a function of beacon and antenna position), is mixed with the complex-valued CW signal $s_{\mathrm{LO}}(t)$ with amplitude A_{LO}, angular frequency ω_{LO}, and phase $\varphi_{\mathrm{LO},k}$ to yield the low-frequency signal

$$\begin{aligned} s_{\mathrm{mix},i,k}(t) &= s_{i,k}(t) \cdot s_{\mathrm{LO}}(t) \\ &= A_{i,k} \cdot s_{\mathrm{TX,BB},k}(t-\tau_{i,k}) \cdot \mathrm{e}^{\mathrm{j}(\omega_{\mathrm{TX}}(t-\tau_{i,k})+\varphi_{\mathrm{B},k})} \\ &\quad \cdot A_{\mathrm{LO}} \cdot \mathrm{e}^{-\mathrm{j}(\omega_{\mathrm{LO}}t+\varphi_{\mathrm{LO},k})} \\ &= A_{\mathrm{mix},i,k} \cdot s_{\mathrm{TX,BB},k}(t-\tau_{i,k}) \\ &\quad \cdot \mathrm{e}^{\mathrm{j}((\omega_{\mathrm{TX}}-\omega_{\mathrm{LO}})t-\omega_{\mathrm{TX}}\tau_{i,k}+\varphi_{\mathrm{B},k}-\varphi_{\mathrm{LO},k})} \\ &\overset{(a)}{=} A_{\mathrm{mix},i,k} \cdot s_{\mathrm{TX,BB},k}(t-\tau_{i,k}) \cdot \mathrm{e}^{\mathrm{j}(-\omega_{\mathrm{TX}}\tau_{i,k}+\varphi_{\mathrm{mix},k})}. \end{aligned} \tag{2}$$

Here, $A_{i,k}$ is the amplitude of the received signal, $A_{\mathrm{mix},i,k}$ is the resulting amplitude, and $\varphi_{\mathrm{mix},k}$ includes the transmitted phase, LO phase, and the phase resulting from $(\omega_{\mathrm{TX}}-\omega_{\mathrm{LO}})t$, which is assumed to be constant for $\omega_{\mathrm{TX}} \approx \omega_{\mathrm{LO}}$ and sufficiently small sample time.

Next, since $\varphi_{\mathrm{mix},k}$ is unknown and a direct evaluation is not possible, the difference between the phases is used to cancel out the unknown term, which is constant for all antenna elements at the receiver. As stated earlier, the phase difference is evaluated by cross-correlation between the ith and the jth antenna as

$$\Delta\varphi_{k,i-j} = \arg\left(\int s^{*}_{\mathrm{mix},i,k}(t) s_{\mathrm{mix},j,k}(t)\mathrm{d}t\right) \tag{3}$$

$$\begin{aligned} &= \arg\Bigg(\mathrm{e}^{\mathrm{j}\omega_{\mathrm{TX}}(\tau_{i,k}-\tau_{j,k})} \\ &\quad \cdot \int s^{*}_{\mathrm{TX,BB},k}(t-\tau_{i,k}) s_{\mathrm{TX,BB},k}(t-\tau_{j,k})\mathrm{d}t\Bigg). \end{aligned} \tag{4}$$

For most practical modulation schemes (e.g., all band-limited baseband signals with bandwidth $\ll f_{\mathrm{TX}}$, pseudo random sequences, or white noise), $\int s^{*}_{\mathrm{TX,BB},k}(t-\tau_{i,k}) s_{\mathrm{TX,BB},k}(t-\tau_{j,k})\mathrm{d}t$ is approximately real-valued and can be omitted for calculation of the phase. This yields

$$\Delta\varphi_{k,i-j} = \mathrm{mod}'_{2\pi}\left(\omega_{\mathrm{TX}}(\tau_{i,k}-\tau_{j,k})\right), \tag{5}$$

where

$$\mathrm{mod}'_{2\pi}(\bullet) = \begin{cases} \mathrm{mod}_{2\pi}(\bullet) & \text{for } \mathrm{mod}_{2\pi}(\bullet) \leq \pi, \\ \mathrm{mod}_{2\pi}(\bullet) - 2\pi & \text{for } \mathrm{mod}_{2\pi}(\bullet) > \pi \end{cases} \tag{6}$$

maps the phase to an interval $(-\pi, \pi]$. (5) is a function of the beacon position, contained in τ and a given receiver position, and there is no remaining unknown parameters. Therefore, multiple measured phase differences from multiple receivers can be processed directly by a Kalman Filter (KF) to compare it with expected phase differences. The beacon position is updated, to minimize the difference between the expected and measured phase differences. A full derivation is given in [26], where a suitable selection of the parameters i and j for choosing antenna combinations is also explained.

V. Experimental Verification

To validate the concept of joint localization and sensing, a practical measurement was conducted, as shown in Fig. 7, using the EMG sensor described in Section III. Three receiver arrays were positioned on the ceiling with approximately 1 m distance in between and 2 m above the measurement area to receive the transmitted data and localize the beacon. Each receiver array employed 12 channels to coherently mix the received RF signal into a signal at the IF of 100 kHz, which was subsequently digitized at a rate of 780 kSPS. In addition, a camera-based tracking system was positioned in the room as a reference, which localized the four reflective markers on the case, as shown in Fig. 6, and calculated the geometrical mean between them, which matched the position of the antenna. The beacon was mounted on the lower arm of a subject, with two EMG electrodes placed approximately 2 cm apart on the musculus deltoideus located on the upper arm. The musculus deltoideus was selected for its easy-to-find innervation zone and its participation in a drawing movement. An elastic strap was attached on one side of a static object, so a drawing on the strap results in a movement in negative y-direction, which is in the area observed by the radar as well as the reference system. Moreover, the drawing force, and therefore, the muscle activity is proportional to the deflection in negative y-direction. This movement was selected to ensure that the transmitting patch antenna always had line of sight to all receiving arrays, which is a requirement and must be addressed for practical applications in further research. During the measurement period of 10 seconds, the participant made two pulling movements of approximately 0.5 m on the band while resting his arm on a box to relax the measured muscle in between.

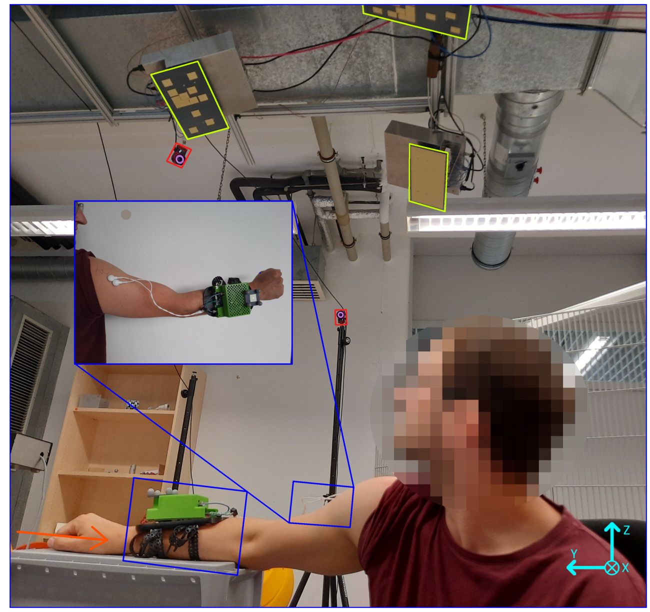

Fig. 7. Experimental validation for simultaneous EMG measurement on the musculus deltoideus and 3D sensor localization. The utilized coordinate system is marked in turquoise, the receiving arrays in green, two of the six cameras of the reference system in red, the sensor with electrodes in blue (which also include a detailed view of the electrode placement), and the pulling direction is indicated by an orange arrow.

Fig. 8 shows the resulting localization using the HEKF compared to the reference from the camera-based system as well as the measured EMG signal. The alignment of the EMG signal with the contraction of the muscle, which is dependent on the

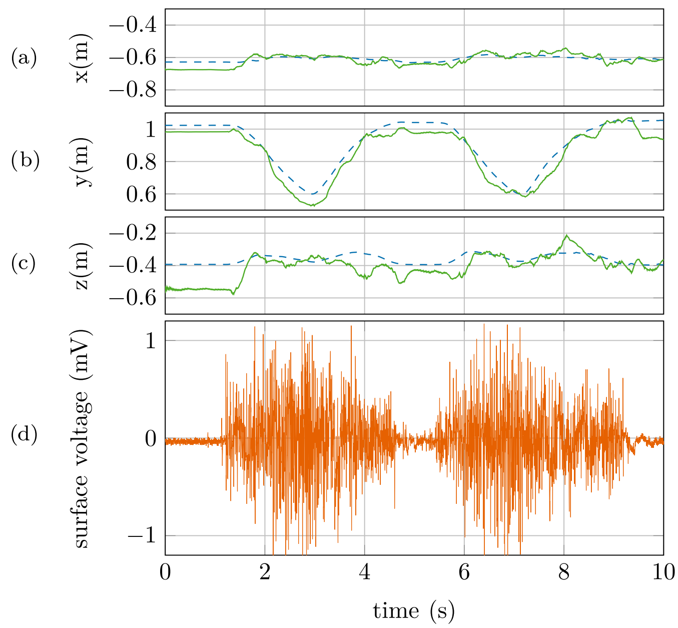

Fig. 8. Results for the conducted measurements, where in (a)–(c), the localization results from the joint localization and sensing system are shown in solid green and the optical reference system as a dashed blue line for the x, y, and z coordinates. Time coherent to that, (d) shows the captured and wirelessly transmitted EMG signal. The deflection in y-direction is correlated to the measured EMG signal, thus revealing the inherent synchronization between them.

force proportional to the displacement in the y-direction, is evident. As stated earlier, the synchronization is inherent to the proposed measurement concept and is not required separately, which is one of the key arguments for the superiority over conventional approaches with multiple different sensors. The root-mean-squared error between the radar and reference for localization is 5.7 cm. This error mainly results from the dilution of precision, in particular in the z-direction, due to the disadvantageous spatial setup, as the entire measurement area had to be covered within line of sight of the receivers and multiple cameras. In [26], a positioning with a root-mean-squared error in the outstanding sub-cm range is shown using the same localization system, where a robotic arm is used as a reference instead of cameras, which reveals the potential of the utilized localization system. Although no medical analysis is conducted in this measurement, it proves the feasibility of the proposed concept and its potential for an all-in-one gait analysis measurement system.

VI. Conclusion

In this article, for the first time, a concept for joint communication and localization EMG-sensing was presented, which has the potential to improve and simplify medical gait analysis. The localization and data transmission were shown to work independently and to not interfere with each other while inherently synchronizing measured data and localization. The example setup demonstrated that this approach does not require a costly hardware and enables broad and inexpensive usage in medical applications. The proposed concept can be expanded to further sensor modalities, such as force sensors or IMUs.

Our concept opens the possibility for new partially automated diagnosis and therapy of diseases such as Parkinson's disease and rheumatoid arthritis.

For an application-oriented implementation, further research should be conducted to connect more base stations for full coverage of the capture area and optimize the beacon in terms of size and power consumption for a sophisticated medical device. While this publication provides a proof for the communication and localization concept, after optimizing the beacon, a medical analysis with Parkinson's disease is the next logical step.

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Stefan Brückner was born in Kronach, Germany, in 1992. He received the M.Sc. degree in electronic engineering from the Friedrich-Alexander-Universität Erlangen-Nürnberg (FAU), Erlangen, Germany, in 2019. He is currently working toward the Ph.D. degree with (FAU). In 2019, he was with the Institute of Microwaves and Photonics, FAU. His research interests include radar for close-range applications, indoor positioning, Kalman filter for localization, and signal processing.

Jasmin Kolpak was born in Bamberg, Germany, in 1987. She received the Dipl.-Ing. degree in electrical engineering, electronics and communication technology from the University of Erlangen-Nuremberg, Erlangen, Germany, in 2011. She is currently working toward the Ph.D. degree in electrical engineering with research on surface EMG sensor topologies with University of Erlangen-Nuremberg. From 2006 to 2011, she studied electrical engineering, electronics and communication technology with the University of Erlangen-Nuremberg, with main focus on microelectronics. Her excellent grades were honored by the Bavarian State Ministry of Science, Research and Art. From 2012 to 2013, she was with MEN Mikro Elektronik GmbH (now duagon Germany GmbH) as the Development Expert for safety-critical hardware applications (SIL 4) in harsh and rugged environments. Since 2014, she has been a Research Assistant with the Institute for Electronics Engineering, University of Erlangen-Nuremberg. Her research interests include the development of biosignal sensors and their biomedical applications as well as innovative telemedical designs.

Fabian Michler (Member, IEEE) received the B.S. and M.S. degrees in electrical and communications engineering from Technische Universität Darmstadt, Darmstadt, Germany, in 2016. In 2022 he received the Dr.-Ing. degree from Friedrich-Alexander-Universität Erlangen-Nürnberg (FAU), Erlangen, Germany, where he is currently leading the Radar Systems team with the Institute for Electronics Engineering. His research interests include high-precision radar systems for vital sign sensing and industrial measurement applications. Mr. Michler was the recipient of the First Prize of the Low-Power Radar Student Design Competition of the IEEE International Microwave Symposium in 2017, 2018, and 2019. He is the Co-founder of the spin-off Sykno GmbH, a company for consulting and research-oriented development of radar systems, RF components, and sensor and communication systems.

Nikita Shanin (Graduate Student Member, IEEE) received the Dipl. degree with distinction in electrical engineering from Bauman Moscow State Technical University (BMSTU), Moscow, Russia, in 2018. Since 2018, he has been working toward the Ph.D. degree with the Institute for Digital Communications, Friedrich-Alexander Universität (FAU) Erlangen-Nürnberg, Erlangen, Germany. His research interests include the broad areas of statistical signal processing and wireless communications, including wireless information and power transfer.

Rober Schober (Fellow, IEEE) received the Diploma (Univ.) and Ph.D. degrees in electrical engineering from Friedrich-Alexander University of Erlangen-Nuremberg (FAU), Erlangen, Germany, in 1997 and 2000, respectively. From 2002 to 2011, he was a Professor and Canada Research Chair with the University of British Columbia (UBC), Vancouver, BC, Canada. Since January 2012, he been an Alexander von Humboldt Professor and the Chair for Digital Communication with FAU. His research interests include the broad areas of communication theory, wireless and molecular communications, and statistical signal processing. He was the recipient of several awards for his work including the 2002 Heinz Maier Leibnitz Award of the German Science Foundation (DFG), the 2004 Innovations Award of the Vodafone Foundation for Research in Mobile Communications, a 2006 UBC Killam Research Prize, 2007 Wilhelm Friedrich Bessel Research Award of the Alexander von Humboldt Foundation, the 2008 Charles McDowell Award for Excellence in Research from UBC, a 2011 Alexander von Humboldt Professorship, 2012 NSERC E.W.R. Stacie Fellowship, 2017 Wireless Communications Recognition Award by the IEEE Wireless Communications Technical Committee, and the 2022 IEEE Vehicular Technology Society Stuart F. Meyer Memorial Award. Furthermore, he was also the recipient of numerous Best Paper Awards for his work including the 2022 ComSoc Stephen O. Rice Prize. Since 2017, he has been listed as a Highly Cited Researcher by the Web of Science. Robert is the Fellow of the Canadian Academy of Engineering, Fellow of the Engineering Institute of Canada, and a Member of the German National Academy of Science and Engineering. He was the Editor-in-Chief of the IEEE Transactions on Communications, VP Publications of the IEEE Communication Society (ComSoc), ComSoc Member at Large, and ComSoc Treasurer. He is the Senior Editor of the Proceedings of the IEEE and as ComSoc President-Elect.

Amelie Hagelauer (Senior Member, IEEE) received the Dipl.-Ing. degree in mechatronics and Dr.- Ing. degree in electrical engineering from the FAU Erlangen-Nuremberg, Erlangen, Germany, in 2007 and 2013, respectively. In November 2007, she was with the FAU Institute for Electronics Engineering, where she researched on BAW resonators and filters toward the Ph.D. degree. Since 2013, she has been focusing on SAW/BAW and RF MEMS components, as well as on microwave integrated circuits for frontends. From 2016 to 2019, she was leading a Research Group on electronic circuits and from August 2019 to September 2021 she was a Full Professor with the University of Bayreuth, Bayreuth, Germany. In September 2021, she was with the Technical University of Munich as a Full Professor and became the Co-Director of the Fraunhofer Institute for Electronic Microsystems and Solid State Technologies EMFT, Munich, Germany. She has authored or co-authored more than 150 peer-reviewed publications in her research fields, which include research and development of microwave theory and technology, electronic circuits and systems, and communication and sensing systems. She was the Guest Editor for a special issue of IEEE Transactions on Microwave Theory and Techniques on the topic RF Frontends for Mobile Radio and as an Associate Editor for IEEE Transactions on Microwave Theory and Techniques.

Robert Weigel (Life Fellow, IEEE) has been Professor for RF Circuits and Systems with the Munich University of Technology, Munich, Germany, from 1994 to 1996. During 1996 to 2002, he was a Full Professor and Director of the Institute for Communications and Information Engineering with the University of Linz, Linz, Austria. In Linz, in August 1999, he co-founded the company DICE, meanwhile split into Infineon Tech-nologies and Apple (in between it has been an Intel company) which are devoted to the design of integrated circuits and systems with more than 400 staff members. Since 2002, he has been a Full Professor and Director of the Institute for Electronics Engineering with the University of Erlangen-Nuremberg, Erlangen, Germany. In 2012, he co-founded the company eesy-ic, Erlangen, devoted to RF and mixed signal circuits with meanwhile more than 60 staff members, since March eesy-ic is a Bosch company. Dr. Weigel has been engaged in research and development of microwave theory and techniques, electronic circuits and systems, and communication and sensing systems. In these fields, he has published more than 1400 papers. He was the recipient of the 2002 VDE ITG-Award, the 2007 IEEE Microwave Applications Award, the 2016 IEEE MTT-S Distinguished Educator Award, the 2018 Distinguished Service Award of the EuMA, the 2018 IEEE Rudolf Henning Distinguished Mentoring Award, and the 2020 IEEE Microwave Career Award. In 2022, his numerous technical and economical achievements were honored with the Cross of Merit First Class of Order of Merit of the Federal Republic of Germany. He is a Fellow of the German ITG, an Elected Member of the German National Academy of Science and Engineering (acatech), and he has been an Elected Member of the Senate of the German Research Foundation (DFG). He is a Member of numerous conference steering and technical program committees. He was the General Chair of several conferences such as the 2013 European Microwave Week, Nuremberg, Germany and Technical Program Chair of several conferences such as the 2002 IEEE International Ultrasonics Symposi-um, Munich, Germany and the 2003 and 2007 EuMA European Conferences on Wireless Technology, both in Munich. He served in many roles for the IEEE MTT and UFFC Societies. He has been the Founding Chair of the Austrian COM/MTT Joint Chapter, Region 8 MTT-S Coordinator, Distinguished Microwave Lecturer, MTT-S AdCom Member, and the 2014 MTT-S President.

Heiko Gaßner studied Sports Science from the Technical University of Munich, Munich, Germany, specializing in the fields of Biomechanics, Medicine, and Exercise interventions. He received the Post-doctoral degree (habilitation) in the quantification of movement disorders and exercise interventions in neurorehabilitation at FAU Erlangen-Nürnberg. In the course of his experimental doctoral thesis with the University of Regensburg, Regensburg, Germany, he evaluated gait characteristics in Parkinson's disease (PD) using Motion Capture and investigated a specific whole-body vibration training for PD patients. In 2014, he accepted a Postdoc position with University Hospital Erlangen and was on several scientific projects using sensor-based technologies to detect motor impairment in neurodegenerative disorders. He was an expert in digital and smartphone-based methods for interdisciplinary care in the home environment. In March 2021, he was appointed the Group Leader for Movement Analysis and Digital Medicine. Since July 2021, he has additionally been the Group Leader for Digital Health Pathways with Fraunhofer IIS. He authored more than 50 peer-reviewed publications in his research fields and is a Member of several international societies focusing on movement analysis: International Society of Posture and Gait Research, European Society for Movement Analysis in Adults and Children, Gesellschaft für die Analyse menschlicher Motorik und ihrer klinischen Anwendung (GAMMA, German-speaking society for movement analysis). In addition, he has a close collaboration with local patient advisory groups of the Deutsche Parkinson Vereinigung.

Jürgen Winkler received the M.D. degree from the Albert-Ludwigs-University of Freiburg, Freiburg im Breisgau, Germany in 1987. He spent his clinical residency with the University Hospital Julius-Maximilians-Universität Würzburg from 1988 to 1992. As a Postdoctoral Fellow he was with the Department of Neurosciences, the University of California San Diego (UCSD), San Diego, CA, USA, under the supervision of Prof. Dr. L. J. Thal, afterwards until 1995 he was a Reserach Fellow with the McDonnel-Pew Center for Cognitive Neuroscience, the Salk Institute,San Diego. From 1995 to 1999, he received the medical board of medicine certificate of the state of California and was appointed as Adjunct Professor of Neurosciences of the Deparment of Neurosciences, UCSD, from 1995 to 2001. In 1999, he was the recipient of the Brookdale National Fellowship Award for Aging (New York). From 2001 to 2007 he was the leading Senior Physician with the University Hospital of Regensburg prior to his Full-Professorship, the Friedrich-Alexander-University Erlangen-Nürnberg, Erlangen, Germany, heading the Department of Molecular Neurology and the Outpatient Center for Movement Disorders. His research focuses on translational research in neurodegenerative diseases. Furthermore, he was a Member of the Governing Board of the Medical School of Medicine, Friedrich-Alexander-University Erlangen-Nürnberg and the University Hospital Erlangen. He was the Speaker of the Bavarian research consortia ForNeuroCell as well as ForIPS from 2006 to 2012 forming the basis for the participation of numerous national and international research consortia focussing on Parkinsons Disease, Huntington Disease and Hereditary Spastic Paraplegia. One of the major clinical focuses is the development of sensor-based methods analyzing gait in movements disorders. His work is reflected by more than 250 peer-reviewed publications. In addition, he serves on multiple advisory boards of neuroscience foundations as well as pharmaceutic companies.

Björn M. Eskofier (Senior Member, IEEE) heads the Machine Learning and Data Analytics (MaD) Lab with the Friedrich-Alexander-University Erlangen-Nuernberg (FAU), Erlangen, Germany. He is also the founding Spokesperson of FAU's Department Artificial Intelligence in Biomedical Engineering, the German Ministry of Economic Affairs and Climate Action GAIA-X usecase project TEAM-X and Co-spokesperson of the German Research Foundation collaborative research center EmpkinS(www.empkins.de). Since April 2023, he has been an Associate Principal investigator and leader of the research group Translational Digital Healthwith the Helmholtz Zentrum Munich. From April 2023 to August 2023, he was a Visiting Professor with Prof. Scott Delp's NMBL lab that is part of Stanford University's Schools of Engineering and Medicine. Dr. Eskofier studied Electrical Engineering with the FAU and graduated in 2006. He received the Ph.D. degree in biomechanics under the supervision of Prof. Dr. Benno Nigg from the University of Calgary Calgary, AB, Canada. He authored more than 400 peer reviewed articles, holds 5 patents, started three spinoff startup companies, and is in a supporting role for further startups. He was the recipient of the several medical-technical research awards, including the Curious Mindsaward 2021 in Life Sciences by Manager Magazin and Merck. In 2016, he was a Visiting Professor with Prof. Paolo Bonato's Motion Analysis Lab, Harvard Medical School (February-March), and in 2018, he was a visiting Professor with Prof. Alex Sandy Pentland's Human Dynamics group, MIT Media Lab (March-August). He was the Area Editor for the IEEE Open Access Journal of Engineering in Medicine and Biology and Associate Editor for the IEEE Journal of Biomedical and Health Informatics. He is also active in the organization of several IEEE and ACM meetings (e.g., BSN, BHI, EMBC, IJCAI, ISWC, UbiComp), most recently as General Chair of BHI 2023.

Martin Vossiek (Fellow, IEEE) received the Ph.D. degree from Ruhr-Universität Bochum, Bochum, Germany, in 1996. In 1996, he was with Siemens Corporate Technology, Munich, Germany, where he was the Head of the Microwave Systems Group from 2000 to 2003. Since 2003, he has been a Full Professor with Clausthal University, Clausthal-Zellerfeld, Germany. Since 2011, he has been the Chair of the Institute of Microwaves and Photonics, Friedrich-Alexander-Universität Erlangen-Nürnberg (FAU), Erlangen, Germany. He has authored or coauthored more than 350 articles. His research has led to more than 100 granted patents. His research interests include radar, microwave systems, wave-based imaging, transponders, RF identification, communication, and wireless sensor and locating systems. Dr. Martin Vossiek is a Member of the German National Academy of Science and Engineering (acatech) and of the German Research Foundation (DFG) Review Board. He is a Member of the IEEE Microwave Theory and Technology (MTT) Technical Committees for MTT-24 Microwave/mm-wave Radar, Sensing, and Array Systems, MTT-27 Connected and Autonomous Systems (as founding chair), and MTT-29 Microwave Aerospace Systems. He is also with the advisory board of the IEEE CRFID Technical Committee on Motion Capture & Localization. Dr. Martin Vossiek was recipient of numerous best paper prizes and other awards. In 2019, he was the recipient of the Microwave Application Award by the IEEE MTT Society (MTT-S) for Pioneering Research in Wireless Local Positioning Systems. Dr. Vossiek has been a Member of organizing committees and technical program committees for many international conferences and was on the review boards of numerous technical journals. From 2013 to 2019, he was an Associate Editor for the IEEE Transactions on Microwave Theory and Techniques. Since October 2022, he has been an Associate Editor-in-Chief for IEEE Transactions on Radar System.