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Busschots, Cedric; Pattyn, Johan; Peumans, Dries; Rolain, Yves; Vandersteen, Gerd

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Forced Oscillation Technique measurement apparatus using fan-speaker hybrid

Cedric Busschots, Member, IEEE, Johan Pattyn, Dries Peumans, Member, IEEE, Yves Rolain, Fellow, IEEE, Gerd Vandersteen, Senior Member, IEEE

Abstract—The number of patients suffering from asthma and Chronic Obstructive Pulmonary Disease (COPD) increases due to deteriorating air quality and smoking. A measurement technique known as the Forced Oscillation Technique (FOT) helps physicians to assess the current state of their patient’s lungs. Measurement devices based on either fans or speakers have been developed in the past that can carry out this measurement, but their performance and/or robustness often fall short. This paper proposes a combination of a fan- and speaker-based measurement apparatus that employs the Forced Oscillation Technique to help physicians diagnose patients with these diseases. First, we discuss the currently available measurement devices and their specific problems. Next, a fan-speaker hybrid measurement apparatus is proposed. This fan-speaker hybrid FOT measurement apparatus is then characterized and its performance is improved with a feed forward compensation. Finally, measurements on two test subjects with the proposed measurement apparatus are shown and discussed.

Index Terms—Medical measurement, frequency-domain measurement, system identification, forced oscillation technique

I. INTRODUCTION

By 2030, Chronic Obstructive Pulmonary Disease (COPD) is projected to be the cause of 7.8% of the total deaths in the world [1]. Already in 2007, the World Health Organization estimated that 210 million people were affected by this disease [2]. The same report mentions that another 300 million people worldwide are suffering from asthma.

In order to diagnose these respiratory patients and check the course of their disease, Pulmonary Function Tests (PFT) have been developed. Of all the available PFT’s, spirometry is the one best known by physicians. A spirometry measurement can be carried out quickly. Due to the high volume of tests and its long service record, comparative data is readily available for this type of test. This offers the physician insight in the state of the patient’s respiratory system. At this time, the diagnosis of COPD and asthma relies heavily on these spirometry measurements [3].

A problem arises when the health condition of a diagnosed patient continues to deteriorate as performing a spirometry measurement is difficult for people with breathing issues. In addition, the spirometric measurements obtained with these patients generally do not offer more insight in the gravity of the illness.

A different non-invasive measurement method, the Forced Oscillation Technique (FOT), has been researched for over 60 years [4]–[6]. Instead of deducting data from the lung capacity and speed of exhalation, an external pressure signal \( p_{\text{gen}}(t) \) is generated and applied to excite the patient’s airway system, while the resulting air flow \( q(t) \) is being recorded. The measurement principle relies on the analogy that can be drawn between air flow and electrical current on the one hand and air pressure and voltage on the other hand. As a consequence, the ratio of air pressure over air flow can be thought of as an impedance. As with spirometry measurements, obstructive airway illnesses such as Chronic Obstructive Pulmonary Disease (COPD) and asthma can be detected and differentiated based on this type of measurement [6]–[8].

The resulting pressure and flow signals from such a FOT measurement can be processed in the frequency domain using the (Fast) Fourier Transform [9]–[11]. In the frequency domain, the total respiratory impedance of the patient is then defined as

\[
Z_{\text{pat}}(j\omega) = \frac{P_{\text{pat}}(j\omega)}{Q(j\omega)},
\]

where \( P_{\text{pat}}(j\omega) \) represents the pressure drop over the total respiratory system. The flow measured at the airway opening is denoted as \( Q(j\omega) \). The complex impedance \( Z_{\text{pat}}(j\omega) \) will be evaluated at every frequency \( \omega \) that is present in the excitation signal. Afterwards, the real and imaginary components of \( Z_{\text{pat}}(j\omega) \) are visualized separately for interpretation by a physician [5].

The Forced Oscillation Technique has been used in a wide frequency range [5]. Implementations of this measurement technique are already in use, both commercially [12], [13] and for research purposes [4]–[6], [13]. These systems operate in a frequency range between 2 and 50 Hz [13].

When low frequencies are contained in the excitation signal, the diagnostic potential of this technique is high. At frequencies below 2 Hz, the viscoelastic behavior of the lung and chest wall are dominant, whereas at frequencies above 2 Hz, the impedance is dominated by the flow resistance of the airway tree and the gas inside the respiratory system [14]. This allows to separate airway and tissue contributions, making the method sensitive to the heterogeneity of alterations across
the lungs [6]. Especially the diagnosis of asthma and COPD benefits from this property [6], [15]. A drawback is that the existing low-frequency measurements require patient training and cooperation with unsedated patients [6]. At present, none of the commercially available solutions make use of this low frequency FOT [13].

Existing literature shows that different types of devices can be used to generate the required pressure signal $p(t)$. A piston exciter is the first way to apply an excitation signal to the patient [4]. Both early and contemporary research often use a speaker system to apply an excitation signal to the patient’s lungs instead. This excitation signal can be applied either at the trachea [6], [16], [17] or from the outside of the chest wall [5]. The bandwidth of these types of systems is high, but such a measurement requires cooperation of the patient [6]. For patients with a respiratory illness, this is not always possible or adequate.

In more recent years, fan-based systems have been developed as well [9], [10], [18]. In contrast to other measurement methods, these types of systems allow a continuous flow of fresh air to the patient. By adapting the excitation signal to the patients breathing, the required level of cooperation of the patient is lowered [11]. As mentioned in [10], [18], [19], the main drawback of this type of setup lies in the limited available bandwidth of 0.1 to 2 Hz due to the low pass behavior of the fan. Therefore, these types of systems only offer low frequency FOT measurements, making the comparison of fan-based measurements with existing FOT literature and devices impossible. In addition, for fan-based systems, the high controller effort required to generate the necessary excitation signal leads to saturation and the related spectral pollution [10], [19].

Expanding the regular Forced Oscillation Technique with low frequencies increases the diagnostic potential without increasing the number of examinations. To overcome the bandwidth limitations of the fan-based setup on the one hand, and to remove the need for patient training on the other hand, a prototype FOT apparatus that combines both fans and speakers has been used. Figure 2 shows a schematic representation of the prototype FOT apparatus that combines both fans and speakers to generate the required pressure signal.

Adding speakers to such a setup by adding an extra opening to which the patient is connected. The flow generated by the fans is perpendicular to the flow to or from the patient. Adding speakers to such a setup by adding an extra opening

Figure 1 shows the electrical analogy mentioned in Section that is used in several reference works [20]–[22]. On this figure, the FOT device is shown as a voltage source, indicated by the blue dotted line. The pneumotachograph can be thought of as a resistance and is shown inside the red dash dotted line. Lastly, the patient acts as a Norton-equivalent current source: the internal impedance $Z_{\text{pat}}(j\omega)$ is the impedance mentioned in (1). The current source $q_{\text{pat}}(t)$ represents the breathing of the patient.

The flow $q(t)$ consists of two parts:

$$q(t) = q_{\text{pat}}(t) + q_{\text{gen}}(t).$$

(2)

$q_{\text{gen}}(t)$ represents the flow generated by the FOT apparatus, whereas $q_{\text{pat}}(t)$ represents the flow caused by the test subject’s own breathing. By using the technique presented in [11], the flow contribution $q_{\text{pat}}(t)$ of the test subject can be spectrally separated from the excitation signal $q_{\text{gen}}(t)$.

The FOT apparatus does not measure flow directly, but rather measures the pressure difference over the pneumotachograph as a measure for that flow. Equation (3) expresses how the flow $q(t)$ can be calculated based on the measured pressures and the resistance of the pneumotachograph.

$$q(t) = \frac{p_{\text{pat}}(t) - p_{\text{gen}}(t)}{R_{\text{ptg}}}$$

(3)

Using this resulting flow $Q(j\omega)$, the impedance $Z_{\text{pat}}(j\omega)$ can be calculated using (1). The real and imaginary parts of $Z_{\text{pat}}(j\omega)$ are then plotted separately. The frequency dependence of the real part changes for test subjects with an obstructive respiratory disease [5]. For patients with COPD, $Z_{\text{pat}}$ tends to increase when compared with healthy test subjects [5]. As a discussion from a physician’s point of view is outside the scope of this paper, the reader is referred to [5] for a more in-depth overview of the diagnostic capabilities of the FOT.

### III. Experimental Setup

In [9]–[11], [18] similar fan-based measurement setups have been used. Figure 2 shows a schematic representation of this type of setup. In these setups, the airflow is created by two fans operating in a push-pull configuration. A T-shaped center piece connects the air flow through the fans with the opening to which the patient is connected. The flow generated by the fans is perpendicular to the flow to or from the patient.
The pressure sensors are highlighted in red. The air flow generated by the fans is perpendicular to the flow to/from the patient.

for the speakers leads to interference between the fans and the speaker.

The hybrid fan-speaker measurement apparatus we propose has separate air channels for the speakers and fans that run alongside each other instead. The two signals are only combined at the pneumotachograph. Keeping the fan signal separated from the speaker signal prevents interference. Figure 3 shows a schematic representation of this setup.

The pneumotachograph is a 3830 series unit manufactured by Hans Rudolph Inc. The usable flow range of this pneumotachograph is 0 to 400 l/min. In [23], this type of pneumotachograph is recommended for Pulmonary Function Tests in adults.

The pneumotachograph has an opening on either side of the membrane. These openings allow to attach a pressure sensor before and after the membrane of the pneumotachograph. Two HCLA12X5EB pressure sensors from SensorTechnics are used to measure these pressures. They have a measurement range of 0 to 1250 Pa, a 12-bit resolution and a response time of 0.5 ms. The symmetric measurement range is necessary since the pressures in the device can be both positive and negative. The response time leads to a bandwidth of 2000 Hz, well above the frequency band of interest.

The pressure sensors are connected to a Microchip PIC32MX795F512L microcontroller using I2C. The microcontroller is mounted on the Ethernet Starter Kit II evaluation board manufactured by Microchip and is attached to a custom Printed Circuit Board (PCB) that interfaces with the sensors, fans, and amplifier. The microcontroller samples the data from the sensors at a sampling frequency of 512 Hz which is more than twice the bandwidth of the amplifier and speakers to prevent aliasing. The same microcontroller feeds both fans and speakers with a Pulse Width Modulated (PWM) signal at the same sample rate. The speed of the fans, and thus the generated pressure, can be controlled directly with such a PWM signal.

The PWM signal for the speakers is filtered by a first order low pass filter to generate an analog signal. This analog signal is fed to the 600 W EcoTec Systems SVC1056-600 class D power amplifier with 0.5% Total Harmonic Distortion (THD). The output of this audio amplifier is coupled to the two speakers in a closed enclosure.

A computer is connected to the microcontroller using the serial port. A Matlab script on the computer controls the measurement setup and gathers the data sent by the microcontroller. The reference signals are generated in Matlab and are sent to the microcontroller as floating point numbers between -1 and 1. The microcontroller converts these to the required PWM values. The internal memory of the microcontroller can hold up to 5120 samples. The electrical part of the setup is schematized in Figure 4.

The fan-based setups used in [10] and [11] can generate pressure signals up to 200 Pa. The amplifier for the speakers allows for a pressure signal that reaches amplitudes higher than 200 Pa. The amplification factor of the speaker system can be adjusted to match the amplitude of the pressure signal generated by the fans. The hybrid fan-speaker setup we propose can generate a Root Mean Square (RMS) pressure of 220 Pa. Momentary peak values of maximum 700 Pa have been recorded with the proposed setup. Although these values are higher than those found in [24], they still remain below the pressures commonly used in mechanical ventilation [25].

### IV. Characterization & Compensation

Before executing measurements on patients, the measurement setup needs to be characterized. A frequency domain model of the measurement setup is determined to this end. This allows to design a feed forward compensation that removes the device characteristic, resulting in an almost ideal excitation at the test subject’s airway opening.

Literature suggests the usage of a random phase multisine excitation when using system identification techniques in the frequency domain [26]. This signal offers a good insight in the frequency response function of linear systems and can reveal the presence of potential non-linear distortion or variations over time [26]. Our multisine excitation signal \( p(t) \) conforms the following expression:

\[
p(t) = \frac{1}{\sqrt{N}} \sum_{k=1}^{N} U_k \sin (2\pi kf_r t + \phi_k)
\]

with \( N \) the number of frequency components, \( U_k \) the amplitude of the frequency component of the \( k \)-th frequency line, \( f_r \) the frequency resolution, and \( \phi_k \) the phase of the \( k \)-th frequency component. Between realizations, the phases \( \phi_k \) vary randomly with a uniform distribution between 0 and \( 2\pi \). In the time domain, these multisines look similar to a Gaussian noise signal when the amplitudes \( U_k \) are chosen identical for each frequency. However, they have the advantage that they are periodic signals. This way, spectral leakage can be prevented.

When applied to the setup discussed in Section III, a frequency resolution \( f_r \) of 0.1 Hz can be obtained. This is due to the sample frequency of 512 Hz and the sample memory of 5120 samples. Therefore, the longest period that can be stored in memory is 10 seconds, leading to a 0.1 Hz frequency resolution.

Due to the frequency resolution of 0.1 Hz, only few frequency lines are available between 0.1 Hz up to 2 Hz. In this
frequency band, all available frequency lines are excited. This is the range of frequencies that is excited almost completely by the fans. Between 2 and 10 Hz, the behavior is less predictable since both the speakers and the fans are actively exciting this frequency region. Only odd frequency lines are used here, as this offers the possibility to detect non-linearities when present [26]. Above 10 Hz, only the speakers will contribute to the pressure signal. Keeping in mind that the FOT only excites frequencies up to 50 Hz, a less detailed characterization is required in the upper frequency band. Therefore, a logarithmic odd frequency spacing has been used for this part of the spectrum. More than one realization of the random phase multisine is needed to characterize the potential non-linearities [26].

At least seven different phase realizations of a random phase multisine signal are necessary to ensure that the statistical asymptotic properties of the estimator used are valid [26, Theorem 10.3, p.387]. For the characterization of this measurement setup, eight phase realizations have been applied. These signals are then applied to the system with the opening for the test subject closed off. By dividing the resulting output spectrum by the input spectrum, a non-parametric transfer function is obtained. This non-parametric transfer function is the Best Linear Approximation (BLA) of the system [26]. This transfer function can be used to approximately match the signal levels of the fans and the speakers by adjusting the volume of the speakers.

Figure 6 shows the BLA transfer function $G_{BLA}$ of the system after matching the volume levels of the fans and speakers. On the same figure, the standard deviation of the combined noise and non-linearity $\sigma_{G_{BLA}}$ and the standard deviation of the noise only $\sigma_{G_{BLA,n}}$ are indicated as well. As can be seen on the figure, the total noise contribution is 20 dB lower than the contribution of the Best Linear Approximation of the system. The presence of non-linearities is revealed by the difference between $\sigma_{G_{BLA}}$ and $\sigma_{G_{BLA,n}}$.

This measurement was also used to obtain a continuous-
A 5 zeros - 10 poles model was selected during the characterization step. This figure shows the pole-zero plot of the selected model.

Fig. 5. A 5 zeros - 10 poles model was selected during the characterization step. This figure shows the pole-zero plot of the selected model.

Fig. 6. shows the non-parametric Best Linear Approximation of the FRF of the measurement system ($G_{BLA}$), represents the standard deviation of both noise and non-linearities ($\sigma_{G_{BLA}}$), represents only the noise component ($\sigma_{G_{BLA},n}$).

The obtained excitation signal is normalized such that the microcontroller again receives a signal between -1 and 1, maximizing the PWM range. Therefore, the gain of the system remains the same as in Figure 6. Figure 7 shows the improved frequency response function of the system when the excitation signal has been shaped with the feed forward controller. This step is carried out before the actual measurement starts. Instabilities in the obtained model do not pose any problems as the periodic excitation signal is known beforehand and all processing happens in the frequency domain.

V. Measurement Results

The measurement apparatus with feed forward compensation has been tested on two human test subjects, one female and one male. A different multisine signal is generated and used as excitation signal. As proposed in [11], the spectral content of the multisine excitation has been adapted to the breathing frequency of the test subjects. This prevents interference between the excitation signal and the breathing of the test subject. Both test subjects were able to comfortably breathe at a breathing frequency around 0.25 Hz. The frequency of the excited lines has been adapted such that they do not interfere with the fundamental breathing frequency or its harmonics. Figure 8 shows the amplitude levels of the adaptive excitation signal that has been used for the measurements with the test subjects. As can be seen here, the frequencies in the lower end of the spectrum have a higher amplitude. This is needed to maintain a high Signal-to-Noise Ratio (SNR) in these frequency bands. The signal level needs to be as high as possible in the low frequency range as the system’s attenuation increases here. On the other hand, both the linearity of the speakers and the patient comfort increase when the signal amplitude of the high frequencies is lowered. Since the SNR is sufficient in the higher frequency range, a lower signal level can be used here without loss of performance.
A similar concept with a decreasing amplitude with increasing frequency was also proposed in [17], [24]. As mentioned earlier, the RMS value of the excitation signal is around 220 Pa at most, with instantaneous values of up to 700 Pa. These values are higher than the ones used in [24] and subsequently recommended in [14], [27]. However, neither of those works uses frequencies below the breathing frequency. In addition, the values applied to the test subjects are lower than those used in mechanically ventilated patients [25].

13 periods of the excitation signal have been applied to the test subjects. The first period is discarded as it is contaminated with transients. To obtain an excitation signal with a frequency resolution of 0.1 Hz, the period length has to be 10 seconds. The total measurement time is therefore 130 seconds, of which 120 seconds can be used.

For the spectral analysis, all 12 periods are processed together, leading to a measurement resolution of 8.33 mHz. As mentioned in [11], it is important that the breathing frequency is a multiple of the measurement resolution. This prevents leakage in the output spectrum of the measurement. In this case, the breathing frequency is 0.25 Hz, a multiple of 8.33 mHz. The relation between the breathing period and the measurement time is fixed by synchronizing the breathing to the measurement.

Different combinations of frequency resolution and measurement time are possible. The frequency resolution selected for these measurements is similar to the frequency resolution offered in [9]–[11].

Figure 9 shows a measurement result for the female test subject. The upper half of the figure shows the pressure \( P(j\omega) \) whereas the lower part of the figure shows the flow \( Q(j\omega) \). In both parts, the excited frequency lines are indicated with circles. The x-marked frequency lines show all other frequency lines on the frequency grid. Both the pressure and the flow spectra show the breathing contribution at 0.25 Hz and the harmonics of this frequency. Due to the synchronization of the test subject to a visual stimulus oscillating at 0.25 Hz, leakage effects remain limited [11]. Therefore, the excitation signal can be clearly distinguished from this breathing contribution. Figure 10 confirms this with a male test subject.

These measurement results show that the proposed fan-sounder hybrid FOT measurement apparatus can apply an excitation signal to the test subject in the frequency band of 0.1 to 50 Hz. The system thus combines the frequency band of 0.1 to 2 Hz of the fan based systems, with the frequency band of 2 to 50 Hz of the commercially available systems by combining the speakers and fans as actuators.

Using (1), the lung impedance can be calculated. To be able to assess the performance of the device when measuring a patient, a different data processing method has been used. The same 12 periods from the earlier measurement are processed in groups of two periods. This leads to a period of 20 seconds. This corresponds to the least common multiple of the period of the excitation signal, 10 seconds, and the breathing contribution, 4 seconds. This way, a frequency resolution of 0.05 Hz can be obtained. By having six blocks of two periods, statistical information can be drawn from the measurement to illustrate its repeatability and variability.
The lung impedance curves are shown in Figure 11 and Figure 13 for test subject one and two respectively. A linear frequency axis was selected to facilitate the comparison to the impedance curves shown in [5].

The uncertainty has been calculated on the basis of the arithmetic mean as it is impossible to calculate measurement uncertainty based on the measurement error of the measurement system. The turbulent flow inside the measurement device is a large contributor to the uncertainty which cannot be expressed analytically. Therefore, we have opted to show only the measurement uncertainty for each measured frequency. For each frequency, the average and standard deviation of the impedance for the six groups of two periods has been calculated. They are shown in boxplot format in Figures 12 and 14 for test subject one and two respectively. Only excited frequencies are shown. The horizontal axes show the index of the excited lines for figure clarity.

Both boxplots show a higher uncertainty in the low frequency range. This behavior can be explained by looking at the amplitudes of the generator signal and the breathing

![Graph 1](image1.png)

Fig. 10. Measurements with a second test subject (male) confirm the functioning of the device.

![Graph 2](image2.png)

Fig. 11. Average impedance curve, calculated from 12 periods of the excitation signal, for the first test subject (female). To facilitate the comparison with similar impedance curves in [5], [12], [13], the frequency axis has been kept linear.

![Graph 3](image3.png)

Fig. 12. A boxplot showing the excited frequencies from the first impedance measurement is shown. The frequency indexes of the excited lines are plotted horizontally for clarity of the figure. The median is indicated as a red diamond, the distance between first and third quartile as a thick blue line. The distance between the minimum and maximum is indicated with a thin blue line. Outliers are displayed as blue circles. The uncertainty on the measurement decreases with increasing frequency. An additional outlier in the resistive part on the first excited frequency line is present at 3014.65 Pa·s/L, but has been omitted from the plot for scaling reasons.
signal. Although there is a clear spectral separation between the breathing and the excitation signal, we suspect that the breathing of the test subject prevents our fan-speaker hybrid from generating an excitation signal that is powerful enough. This effect is the most pronounced in the lowest part of the frequency band. Increasing the amplitude of the excitation signal might be option. However, care should be taken not to damage the respiratory system of the patient. To the authors’ knowledge, no conclusive upper pressure limits for FOT measurements have been published up to now.

A. Comparison with other FOT devices

References [12] and [13] show that a direct comparison of the values of the impedance measured by commercial devices reveals differences between them, even when measuring artificial and normalized loads. However, the shape of the impedance curves in Figures 11 and 13 is similar to those shown in [5] and [13]. For both the reference measurements and the measurements with the fan-speaker hybrid, a constant value is obtained after 20 Hz. Depending on the measurement apparatus, a value between 240 to 300 $\frac{\text{Pa} \cdot \text{s}}{\text{L}}$ for male test subjects and between 250 and 360 $\frac{\text{Pa} \cdot \text{s}}{\text{L}}$ for female test subjects was obtained [13]. In the case of the female test subject shown in Figure 11, the resistances found above 20 Hz range from 349 to 381 $\frac{\text{Pa} \cdot \text{s}}{\text{L}}$. For the male test subject, values between 185 and 254 $\frac{\text{Pa} \cdot \text{s}}{\text{L}}$ have been obtained.

Reference [13] also shows that the sign of the reactance changes between 10 and 15 Hz. This behavior can be seen as well in Figure 13 for the male test subject. In the case of the female test subject, the mean reactance measured with the fan-speaker hybrid measurement apparatus shows a sign change at a frequency above 15 Hz.

The values of the resistance and reactance are close to the values reported by [13], but do not match them completely. The test subjects measured with the fan-speaker hybrid measurement apparatus are substantially younger and have a lower Body Mass Index than the average participant in [13]. Both parameters have an important influence on the lung impedance [13].

To the authors’ knowledge, no measurement data showing the measurement uncertainty is available for fan-based systems. However, the Signal-to-Noise Ratio (SNR) of the flow in frequencies below 5 Hz is equal to the SNR found in [11]. In addition, the values for the pressure and flow signals have the same order of magnitude as those reported in [11].

VI. Conclusion

A Forced Oscillation Technique measurement apparatus that combines a fan-based and a speaker-based actuators has been developed. Instead of choosing between a low frequency FOT measurement (0.1-2 Hz) or a regular FOT measurement (2-50 Hz), a single measurement cycle is now sufficient to obtain a measurement that combines the diagnostically interesting low frequencies with the state-of-the-art FOT measurement techniques. The proposed measurement device does not require patient-unfriendly maneuvers from the test subject, as he/she can continue to breathe throughout the measurement cycle while fresh air is being provided continuously. The fan-speaker hybrid measurement apparatus has been tested on two human test subjects. In both cases, the breathing signal could be separated from the excitation signal. The resulting impedance curves match the ones shown in literature, but the uncertainty on these curves in the low frequency part of the spectrum remains higher than the uncertainty in the higher frequencies.
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Cedric Busschots was born in Leuven (Belgium) in 1990. In 2012, he obtained the degree of ‘industrieel ingenieur in de elektronica-ICT’ at Lessius Mechelen - Campus De Nayer. He continued at the VUB and ULB, and graduated in 2015 with a degree in electromechanical engineering. In August 2015, he joined the ELEC department as a PhD student. His research focuses on system identification using microcontrollers.

Johan Pattyn was born in 1974 (Belgium). After 2 years of engineering studies, he joined the navy to become a radar technician. Since December 2009 he is with the department ELEC of the Vrije Universiteit Brussel (full time tenure). He is responsible for the Rapid PCB Prototyping and is also involved in the maintenance and repair of the instruments and circuits for the students lab’s.

Dries Peumans (1992, Belgium) received the M.Sc. degree and the Ph.D. degree in Electrical Engineering (Electronics and Information Technology) from the Vrije Universiteit Brussel (VUB), Belgium, in 2015 and 2020, respectively. His current research interests include the analysis, modeling and measurement of complex (non)linear systems in the frequency domain.
Yves Rolain (1961, Belgium) received the Electrical Engineering (Burgerlijk Ingenieur) degree in July 1984, the degree of computer sciences in 1986, and the PhD degree in applied sciences in 1993, all from the Vrije Universiteit Brussel (VUB), Brussels, Belgium. He was a research professor at the VUB in the department ELEC, and since 2014 he is a full-time professor in electrical engineering at the same department. He became a fellow of the IEEE in 2006 and received the IEEE I&M Society award in 2005. His main interests are microwave measurements and modeling, applied digital signal processing and parameter estimation/system identification.

Gerd Vandersteen was born in Belgium, in 1968. He received the degree in electrical engineering and the Ph.D. degree in electrical engineering from Vrije Universiteit Brussel (VUB), Brussels, Belgium, in 1991 and 1997, respectively. He was with the Micro-Electronics Research Centre IMEC, Wireless Group, as a Principal Scientist, with a focus on modeling, measurement, and simulation of electronic circuits in the state-of-the-art silicon technologies. This research was in the context of a collaboration with VUB. Since 2008, he has been a Professor with the Department of Fundamental Electricity and Instrumentation, VUB, within the context of measuring, modeling and analysis of complex linear and nonlinear system. Within this context, the set of systems under consideration is extended from micro-electronic circuits toward all kinds of electro-mechanical systems. Since 2011, he has been the Director of the Doctoral School of Natural Sciences and (Bioscience) Engineering (NSE), VUB.