TECHNICAL BULLETIN

SenTec EIT: Principle of Operation[1]

senTec
Advancing Noninvasive
Patient Monitoring

[1] This document is referring to the current state of the art and SenTec software versions TIC 1.6.x.xxx / GUI 1.0.x.x, where “x” can be any number. Future changes of the aforementioned software versions may prompt adaptation of this document.

Contents

Introduction

SenTec EIT is based on the principles of chest electrical impedance tomography (EIT), a radiation-free technology that offers the possibility to continuously monitor lung function and ventilation distribution. SenTec EIT is available in two configurations, one for adults/children and one for neonates/infants, each with specifically configured software but with the same underlying technology.
This document intends to summarize the main concepts, the principle of operation and limitations that underpin SenTec EIT. For more information on EIT in general we refer e.g. to the consensus paper by Frerichs et al.[2] and to the book chapter on EIT in Eckart et al.[3], which were also used as reference sources for some parts of this document.

Measurement concept

The patient to be examined wears a disposable textile belt (Figure 1) with 32 electrodes embedded in a structured fabric.

Figure 1: Neonatal EIT belt.

The belt is designed to follow the ribs in order to avoid any impairment of the ribcage movement that might increase the work of breathing. As the ribcage anatomy is age-dependent, the belt follows the ribs on a transverse plane in neonates (Figure 2) and an oblique plane in adults (Figure 3).

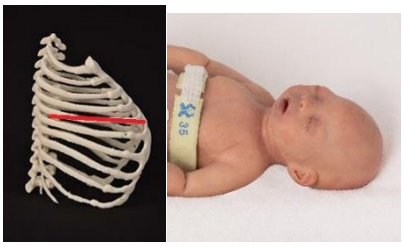

Figure 2: Illustration of a neonatal ribcage[4] and a belt placed around the chest on a transverse plane.

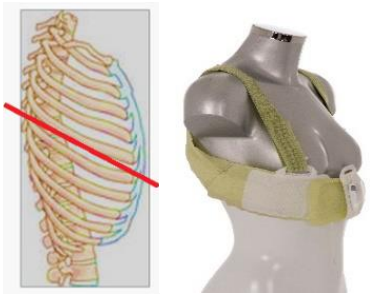

Figure 3: Illustration of an adult ribcage and a belt placed around the chest on an oblique plane.

Very weak alternating electrical currents are then applied in turn to pairs of electrodes (Figure 4) while the voltages resulting from the electrical field are measured on pairs of the remaining electrodes. There are 32 pairs of electrodes the electrical current is applied to in turn. For each of these pairs there are 32

[2] Frerichs et al., Chest electrical impedance tomography examination, data analysis, terminology, clinical use and recommendations: consensus statement of the TRanslational EIT developmeNt stuDy group, Thorax 72(1), 2017, pp. 83-93.
[3] Hammermüller et al., Möglichkeiten und Grenzen der modernen elektrischen Impedanztomographie (EIT), in Eckart et al., Intensivmedizin, ecomed MEDIZIN 2016, pp. 1-19.
[4] Image adapted from http://yffverstraeten.nl/ribben.html.

measurements of the voltages through 32 pairs of electrodes. Overall there are 32 times 32 i.e. 1024 measurements to build a frame, which is the base of one tomographic image. With the actual system, the temporal resolution is about 50 frames per second. This sampling rate is much faster than the respiratory and heart rates, so that the derived signals have low distortion and can be visualized with good resolution in the time domain. However for very high-frequency signals, for example in the range around 10 to 11 Hz as typically encountered under high-frequency oscillation ventilation (HFOV) only a simplified type of analysis is currently available.

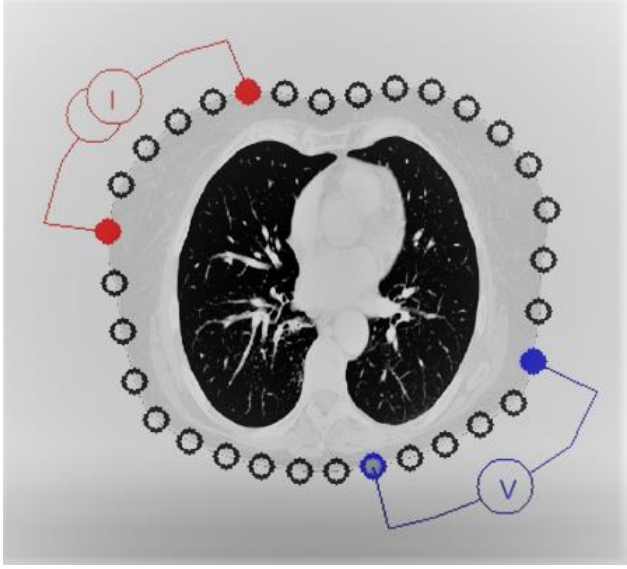

Figure 4: Schematic drawing of the electrode locations on the chest circumference, with electrical current being applied to a pair of electrodes (in red). The current applications are successively shifting around the chest. For each current applied to a pair of electrode, the voltages are measured by 32 pairs of electrodes.

The EIT measurements allow to derive information on the distribution of electrical impedance within the thorax using algorithms specifically developed for ill-posed non-linear problems. Without going into much detail, these are mathematical problems which do not admit a unique solution, so in order to overcome them, the algorithms make use of additional assumptions such as smoothness of the intra-thoracic impedance distribution or a-priori information such as models of thorax and lung contours. Moreover another simplification consists in focusing on changes of the impedance distribution over time.
The voltage is directly proportional to the impedance, which is basically the resistance that the current encounters in its path. In formulas, according to Ohm's law: V= R×I, with V voltage, R resistance and I current. The impedance is the analog of R when the current is alternating.
When the patient under examination inspires and air fills the lungs, the lung tissue stretches, so that the current goes through thinner and longer paths, with higher overall resistance. A larger ventilation and stretch in a specific lung area can then be related to higher impedance values measured in the corresponding region on the thorax, so that the image which is then reconstructed can show with pixel-wise resolution the lung distribution of impedance values.
Therefore if we follow inspirations and expirations, as in Figure 5, we can expect to measure corresponding changes in impedance, increasing with increasing lung volumes and vice versa. For example the *End-Expiratory Lung Impedance* (*EELI*)[5] is related to the functional residual capacity (FRC) during quiet tidal breathing, or to the residual volume (RV) after a forced full expiration (see Figure 5 for a representation of FRC and RV).
Since intra-thoracic impedance varies with ventilation, the recorded impedance changes can be used to generate clinically relevant information about ventilation distribution in form of waveforms, real-time tomographic images or indices. The arrangement of the electrodes and the pattern of current injections enable a resolution that allows the evaluation of ventilation and related lung volume changes even on a local or regional level.
It should be noted that there are other factors that may contribute to impedance changes such as those related to heart action and perfusion, but generally they are to a great extent negligible.

[5] As the nomenclature is not always completely consistent among different EIT devices, italic capital letters are used to identify SenTec specific terms and definitions. In the section 'Information Display' of this document the main terms as used by SenTec are explained.

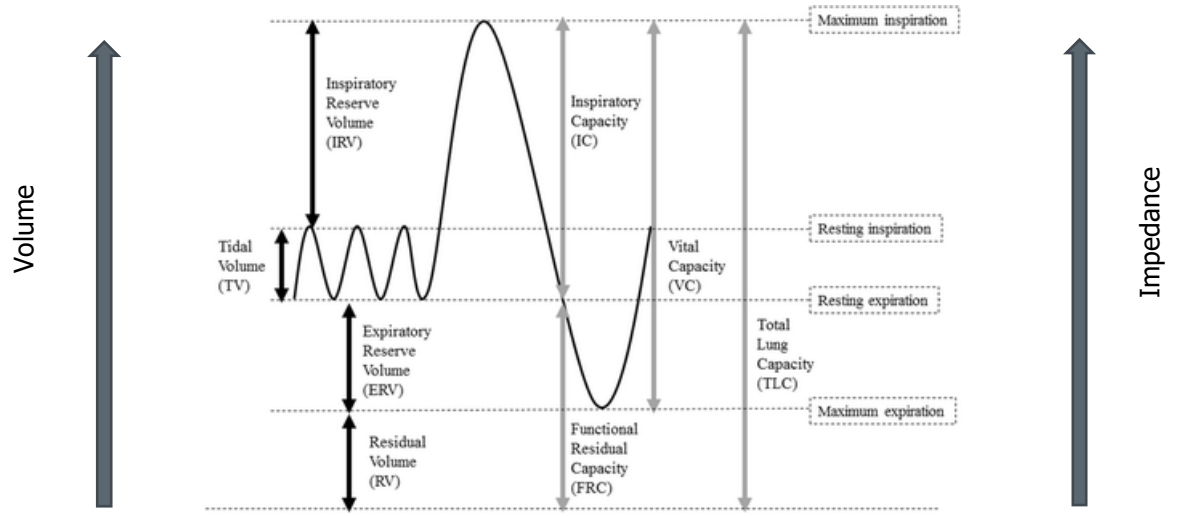

Figure 5: Representation of lung volumes during quiet tidal breathing and forced full inspiration and expiration[6]. As lung volume increases, the measured impedance increases as well, and vice versa.

Moreover, it is worth keeping in mind that what really matters in the current EIT paradigm is how the impedance values change over time rather than the absolute values of the measurements. For example FRC is an interesting indicator that is not measurable with a common spirometry; EIT in the current implementation is not able to measure FRC at a given time point, but can nevertheless monitor how FRC changes over time on a global as well as on a local or regional level.
It should also be remarked that, although we talk about impedance, the values obtained through the reconstruction algorithms are expressed in arbitrary units, as they are not units of a physical variable.

Particularities of SenTec EIT

The SenTec EIT system consists of a monitor with a *Tomographic Image Creation* (*TIC*) module, a belt connector, a textile belt, and a contact medium to be applied between belt and skin.
The data acquired through the measurements undergo a processing stage that leads to output information. Each step in the process from data acquisition till displayed output contributes in making each EIT device unique. Among specific features of SenTec EIT the following need to be mentioned:

- The reconstruction algorithms and data processing make use of models of thorax and lung contours derived from computed tomography (CT). This allows to contextualize measurements with respect to individual patients because characteristics such as weight, height, gender, chest circumference are taken into account in the models (Figure 6).

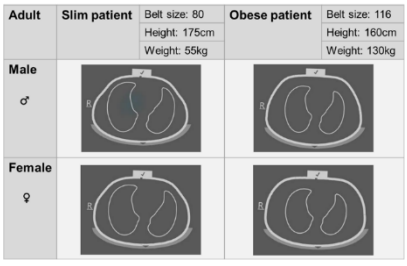

Figure 6: Models of different thorax and lung contours

This approach only focuses on data that is related to the lung ventilation distribution and displays reconstructed images within the *Lung Contours* for a better interpretability and contextualization. This in turn allows the introduction of novel indicators or concepts, such as *Silent Spaces*, as explained in the next section. Also in terms of accuracy, it has been shown that anatomically enhanced EIT-image reconstruction is superior to non-individualized reconstruction[7].

[6] Image adapted from: Lutfi, The physiological basis and clinical significance of lung volume measurements. Multidiscip. Respir. Med. 12(3), 2017, pp.1-12.
[7] Thürk et al., Effects of individualized electrical impedance tomography and image reconstruction settings upon the assessment of regional ventilation distribution: Comparison to 4-dimensional computed tomography in a porcine model. PLoS ONE 12(8), 2017, pp.1-16.

- A sensor, which may be embedded either in the belt or in the connector, continuously detects the patient position, whether prone, supine, lateral or inclined, which also allows for a better interpretation of data, since gravity significantly influences lung mechanics and ventilation distribution.

- The belts feature an adhesive-free, textile belt/skin interface, which is particularly important for patients with sensitive and fragile skin such as preterm neonates. They are designed to ensure a tight fit without restricting breathing, by following the movement of the ribs, as explained above, and thanks to an expandable closure band in the neonatal belts. These design features as well as the materials minimize the risk of skin lesions, while the single-patient use of the belts prevents cross-contamination. These advantages as well as good performance and contact quality are documented in literature [8,9,10].

- The contact quality between skin and electrodes is enhanced by a skin-friendly contact medium applied between skin and belt. Moreover contact quality is continuously assessed, which helps to infer when measurements are more or less reliable, and in case of impaired contact of up to 6 electrodes, an automatic compensation can still safeguard the quality of the signal.
  Further, to avoid an inadvertent belt displacement around the patient's thorax, shoulder strips are added to the belts in the adults/children setup, whereas for neonates/infants the ability to account for inadvertent belt displacement is included directly in the calculations (provided this is measured by the operator and entered as input via GUI).

Information display

First of all, the monitor displays *Global Dynamic Images* (Figure 7) showing impedance changes within the cross-section of the thorax in real-time.

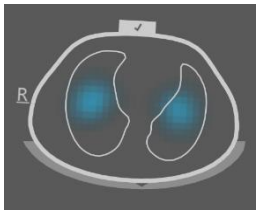

Figure 7: Example of a *Global Dynamic Image*: brighter colors are related to larger impedance change.

The color changes related to impedance variation can be put in relation with the respiratory cycles (Figure 8) where brighter colors point to a larger impedance change. To enable a better color visualization, impedance changes are plotted with respect to a moving reference frame. Also, the *Lung Contours* are superimposed for a better visualization and interpretation.

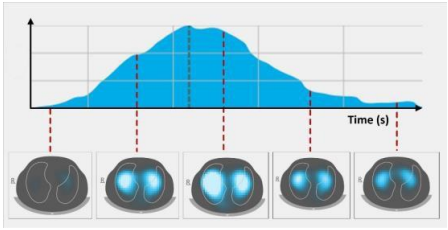

Figure 8: Example of one breath cycle where impedance changes over time are put in relation to a sequence of *Global Dynamic Images*. The diagram above shows the *Lung Impedance* waveform, referred to as *Plethysmogram*, representing changes over time of the total impedance within the *Lung Contours*.

Differently from the *Global Dynamic Image*, all other images and indices are derived by focusing only on the pixels within the *Lung Contours* (thereafter referred to as *Lung Pixels*).

[8] Waldmann et al. Performance of Novel Patient Interface for Electrical Impedance Tomography Applications. J. Med. Biol. Engin. 37, 2017, pp. 561–566.
[9] Sophocleous et al. Clinical performance of a novel textile interface for neonatal chest electrical impedance tomography. Physiol. Meas. 39(4), 2018, pp.1-11.
[10] Becher et al. Feasibility and safety of prolonged continuous monitoring with electrical impedance tomography in neonates and infants with respiratory failure. Intens. Care Med. Exp. 7(3), 2019, pp. 209-210.

The *Plethysmogram* (Figure 8) is a waveform representing how the *Lung Impedance*, i.e. the sum of the impedance values of all *Lung Pixels*, changes over time, thereby reflecting lung volume variations with breathing.

The *Plethysmogram* allows the identification of individual breath cycles, which can be used to derive a respiratory rate based on *Lung Impedance* changes. The corresponding algorithm involved in the calculation of the respiratory rate is intended to only display reliable values. Therefore, in certain situations where the respiration pattern is quite irregular in frequency and/or presents significant amplitude variations, no value is provided.

Indeed, the user is offered three analysis modes to encompass also those cases where breath detection is impaired and ventilation is irregular. Besides the *Breath-by-Breath* (*BB*) mode, there are two *Time-Based* modes (*TB-I* and *TB-II*) that the user can choose from for a better data interpretation. To explain this concept, three illustrative examples of breathing patterns, artificially constructed for educational purpose, are considered in Figure 9.

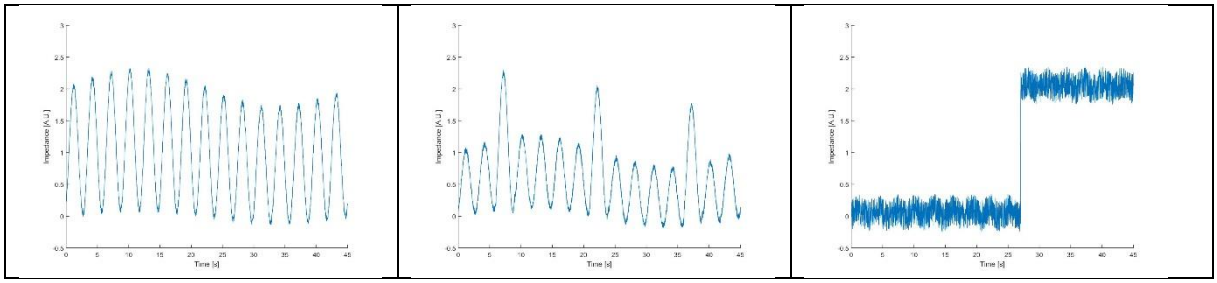

Figure 9: Simulation of three breathing patterns

• To analyze breathing patterns that are quite regular, as in the first example, basically when most breaths can be detected reliably (e.g. mechanically ventilated patients or spontaneously breathing patients with regular breathing patterns characterized by regular respiratory rates and tidal volumes), the *BB* mode is recommended.
• To analyze breathing patterns, as in the second example, that present significant variations in amplitude and/or frequency, which may hinder a reliable breath detection (e.g. spontaneously breathing patients with irregular breathing rate and/or varying tidal volumes, or patients breathing rather shallowly), the *TB-I* mode is recommended.
• With no clear breathing patterns, as in the third example, usually with very low tidal volumes when cardiac-related impedance changes are large compared to ventilation-related impedance changes (e.g. patients being ventilated at high breathing frequencies such as HFOV or patients on extracorporeal life support), the *TB-II* mode is recommended.

With the *BB* mode the end of expiration and end of inspiration for each breath are identified in the *Plethysmogram* (Figure 10). The corresponding values of *Lung Impedance* at these time points represent the *End-Expiratory Lung Impedance* and *End-Inspiratory Lung Impedance* (*EELI* and *EILI*).

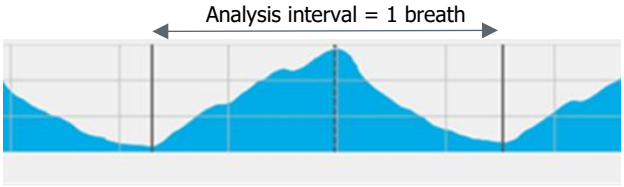

Figure 10: Example of a *Plethysmogram* where the solid vertical lines identify the end of expiration and the dotted one the end of inspiration. The corresponding impedance values are the *End-Expiratory Lung Impedance* (*EELI*) and *End-Inspiratory Lung Impedance* (*EILI*).

The difference between *EILI* and the preceding *EELI* is the *Tidal Variation*, which is related to the tidal volume inhaled in one breath. The monitor displays the trend data of *EELI* and *EILI*, so that changes over time of the corresponding lung volumes can be followed. To derive the trend, consider that each point in the *EELI* trend corresponds to a point in the *Plethysmogram* identified by a solid vertical line (Figure 10) and each point in the *EILI* trend corresponds to a point identified by a dotted vertical line (Figure 10). This is shown schematically in Figure 11.

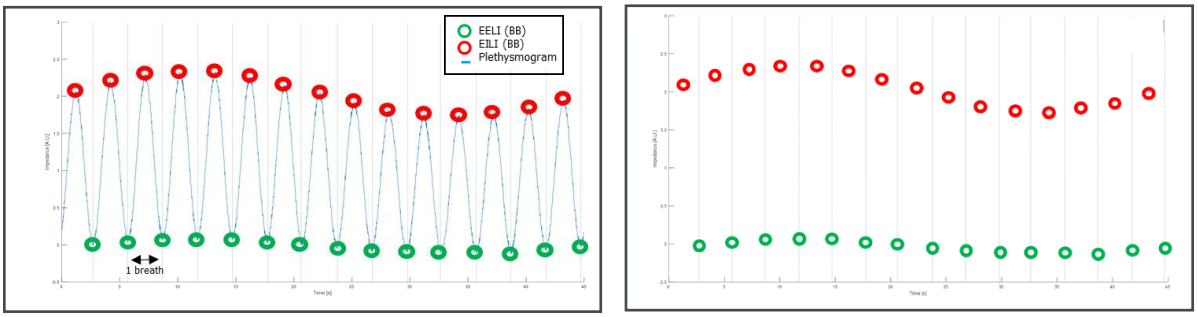

Figure 11: Schematic drawing of *EELI* trend in the *BB* mode: on the left the *Plethysmogram* with *EELI* and *EILI* points highlighted as circles; on the right only the trends.

In the *TB-I* mode the analysis is similar, but made on intervals of 15 seconds: the minimum and maximum of *Lung Impedance* are identified in each interval and still called *EELI* and *EILI*, although not necessarily referring to a single breath cycle (Figure 12).

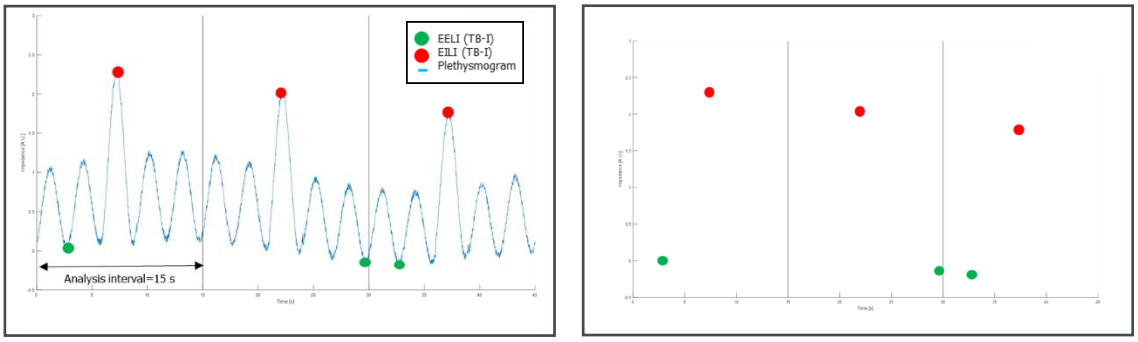

Figure 12: Schematic drawing of *EELI* trend in the *TB-I* mode: on the left the *Plethysmogram* with *EELI* and *EILI*; on the right only the trends.

In the *TB-II* mode the analysis is still made on intervals of 15 seconds, but *EELI* and *EILI* are not shown in the trend chart anymore. Instead the trend chart shows the *Aeration* as the mean of *Lung Impedance* values over each 15 second interval (Figure 13). As such, Aeration represents the mean lung volume in each interval.

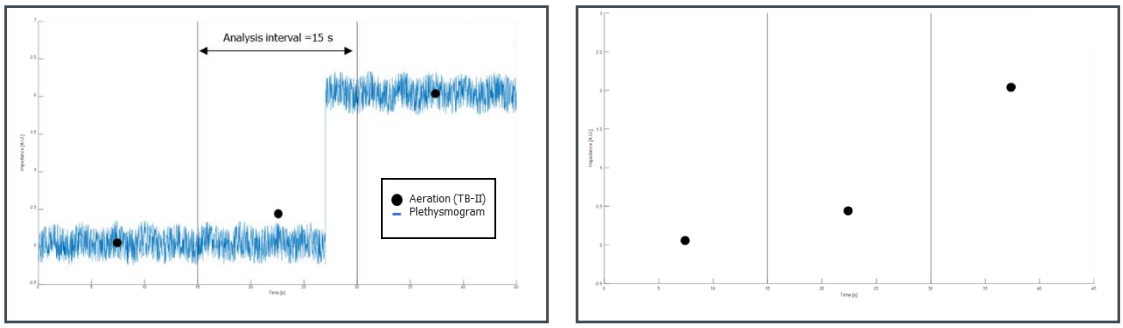

Figure 13: Schematic drawing of *Aeration* trend in the *TB-II* mode: on the left the *Plethysmogram* with *Aeration* values; on the right only the trends.

It is important to note that each mode is available for any signal, it is up to the user to choose what makes most sense for the interpretation. Of course an irregular signal may not yield any output in the *BB* mode, depending on how breaths are detected. Further, the *BB* and *TB-I* modes identify a maximum and a minimum and, based on that, enable the derivation of further images and indicators on the ventilation distribution. Depending on the shape of the *Plethysmogram*, however, the interpretation of the displayed output may not always be straightforward, therefore certain breathing patterns are better analyzed in the *TB-II* mode.

We can use our 3 example waveforms to exemplify how to compare the possible outputs (Figure 14).

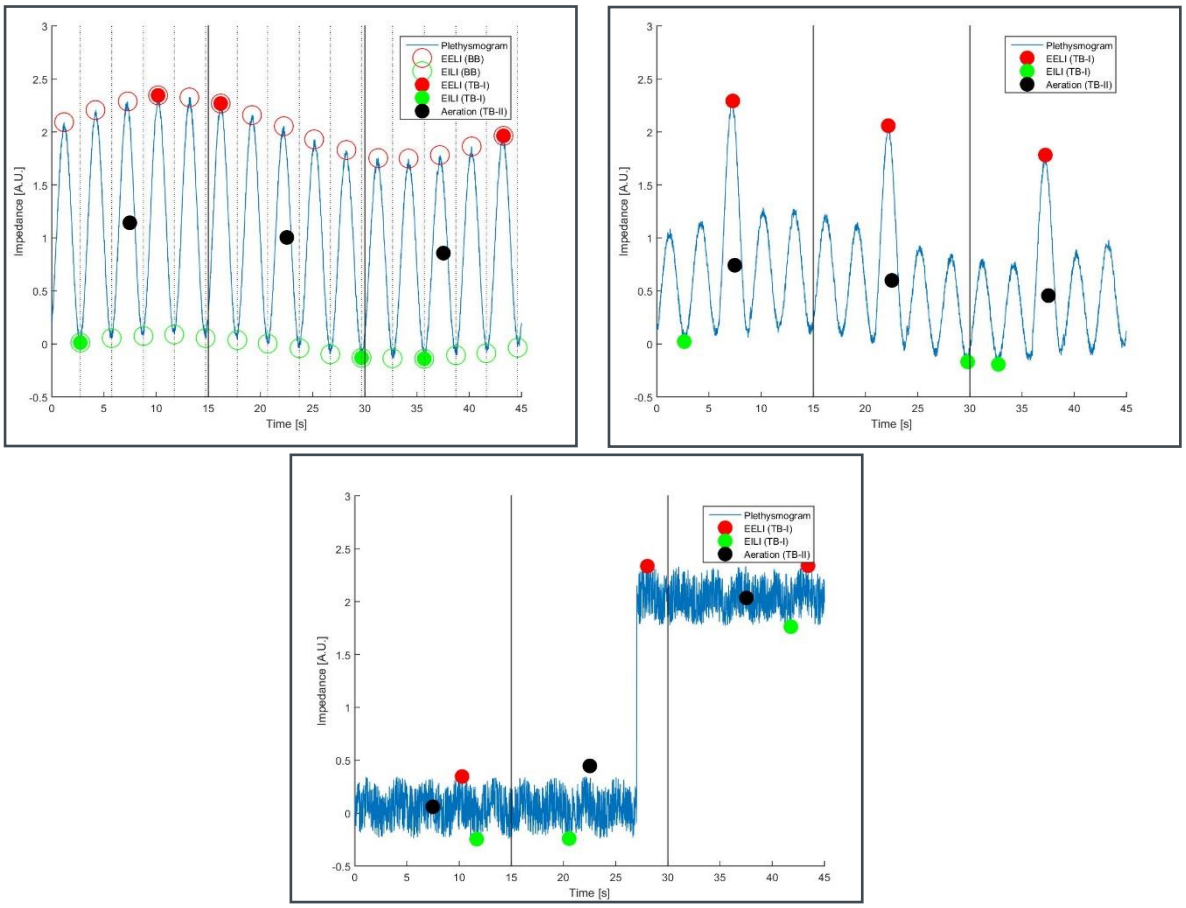

Figure 14: Schematic drawings of trend in different analysis modes for the three example waveforms. *EELI* and *EILI* in the *BB* mode are only shown in the first example where we assume breaths are detected reliably.

While in the *TB-II* mode the analysis is limited to *Aeration* only, in the *BB* and *TB-I* modes, several other indicators and images can be derived by comparing the maximum and the minimum values within the analysis interval.

First of all, for every analysis interval (a breath or 15 seconds depending on the mode), it is possible to compute for each *Lung Pixel* its impedance change between the two time points representing *EELI* and *EILI*. This is like computing the difference between images at the two time points (Figure 15).

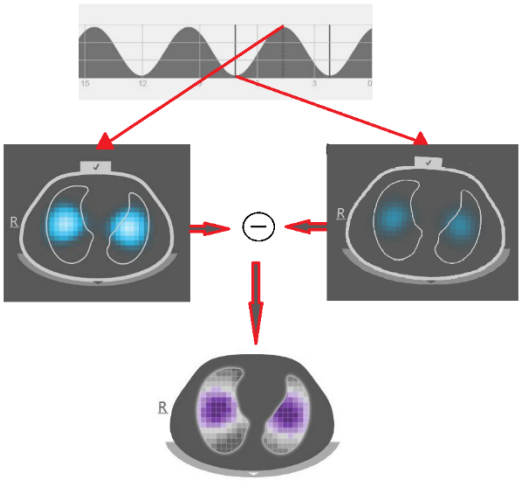

Figure 15: Schematic illustration of the concept of *Stretch Image*: the image on the left represents ventilation distribution measured at the end of inspiration and the image on the right represents ventilation distribution at the end of expiration. The 'subtraction' process is explained in the text.

To be more precise, the *Lung Pixel* with the maximum change is identified and taken as the 100% reference for the following definition: the *Relative Tidal Stretch* (*RTS*) of a *Lung Pixel* is defined as its impedance change expressed in percentage of the maximum impedance change of all *Lung Pixels*.
Assigning a value to each *Lung Pixel* using this approach leads to the so called *Stretch Image*, where these values are represented in color scale (Figure 16) to convey a visualization of ventilation distribution.

Figure 16: Example of a *Stretch Image*.

Regions with large changes, reflecting better ventilation, are shown in violet colors. Regions with small changes are displayed in greyish colors.
To visualize how much different regions are contributing to the total lung impedance change between the two identified time points, a ten-part bar chart is displayed (Figure 17). Each bar represents a cluster of pixels with similar *RTS*, whereby the first bar shows those pixels with *RTS* between 0% and 10% and the last bar those with *RTS* between 90% and 100%. The height of each bar is the percent contribution of the cluster in the total lung impedance change.

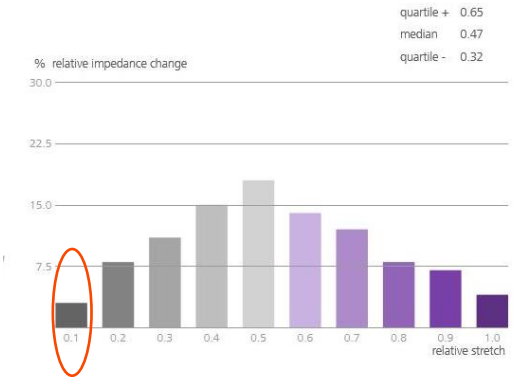

Figure 17: Example of a *RTS* ten-part bar chart. The first bar corresponds to the *Silent Spaces*.

The pixels belonging to the first bar constitute the so-called *Silent Spaces*, representing those regions within the lung that exhibit little or no ventilation. The rest of the pixels constitute the so-called *Functional Lung Spaces* (*FLS*), related to the concept of Functional Lung Size[11].
From the *Stretch Image* it is also possible to geometrically determine a *Center of Ventilation* (*CoV*) by weighing each pixel *RTS* value. So the *CoV* can be defined as the weighted geometrical center of ventilation distribution within the *Lung Contours*. The coordinate *CoV(v)* characterizes the ventilation distribution in the vertical direction, along the gravity vector, whereas *CoV(h)* characterizes the ventilation distribution in the horizontal direction. The horizontal line through *CoV*, perpendicular to the gravity vector, is called the *Horizon of Ventilation* (*HoV*). For the sake of comparison, another center of ventilation is also derived assuming homogeneous ventilation in the lungs, i.e all weights being the same; we refer to it as *Center of homogeneous Ventilation* (*CoV*^{hom}). These concepts are illustrated in the Silent Spaces Image (Figure 18) discussed below.

[11] Amato et al. Driving Pressure and Survival in the Acute Respiratory Distress Syndrome. New Engl. J. Med. 372(8), 2015, pp. 747-755.

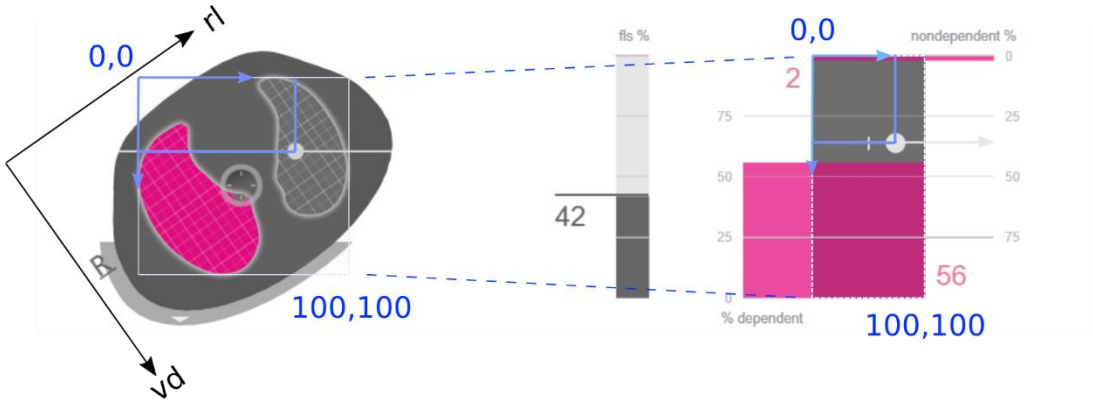

Figure 18: Example of *Silent Spaces Image*, *Functional Lung Spaces* (*FLS*) bar and *Silent Spaces* bar. The white dot and grey hollow circle represent the *Center of Ventilation* (*CoV*) and the *Center of homogeneous Ventilation* (*CoV^{hom}*). Two coordinate systems are drawn for explanation purpose, one in blue with respect to gravity, and the other in black with respect to the patient (*rl* for right-left, *vd* for ventrodorsal direction)

The *Silent Spaces* below the *Horizon of Ventilation* are called *Dependent Silent Spaces* (*DSS*). The *Silent Spaces* above the *Horizon of Ventilation* are called *Non-Dependent Silent Spaces* (*NSS*).
This subdivision is meaningful insofar as gravity influences the ventilation distribution and may point to a different pathophysiology, e.g. *DSS* may be due to collapse or fluid accumulation in alveoli while *NSS* may rather be due to over-distension. If the patient is standing or sitting upright and the examined thorax cross-section is basically on a horizontal plane, then the subdivision is not so meaningful because *DSS* and *NSS* are on the same level with respect to the gravity vector. It is worth mentioning that the monitor continuously shows rotation and inclination of the patient.
It is important to note that, thanks to the position sensor, the coordinates *CoV(v)* and *CoV(h)* are always automatically defined, independently of body position, in relation to the orientation of the gravity vector, allowing to directly assess the influence of the patient's position on the ventilation distribution.
If the patient is in supine position, these coordinates are aligned with the coordinates defined by the patient's main body directions, namely the right-left and ventrodorsal coordinates, *CoV(rl)* and *CoV(vd)* respectively. If the patient is standing or sitting upright, CoV(v) and CoV(h) are not very meaningful.

The *FLS* bar (Figure 18) represents the percentage of *FLS* with respect to the total lung area, and the *Silent Spaces* bar (Figure 18) represents the respective percentages of *NSS* and *DSS* (*FLS*, *DSS* and *NSS* together totaling 100%). Here *CoV* is also displayed for contextualization, with its vertical coordinate represented on the vertical scale (the horizontal one being also displayed but without a scale). These percentages and the vertical coordinate of *CoV* are also shown in the corresponding trend graph (Figure 19).

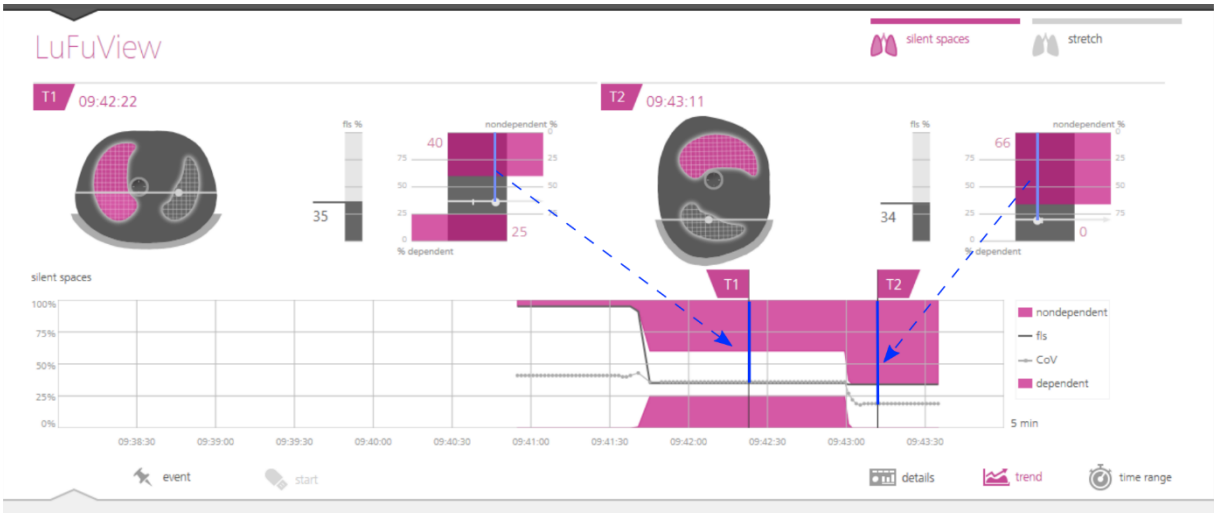

Figure 19: Example of a *Silent Spaces* trend graph. Two time points, T1 and T2, are set for comparison. *Dependent Silent Spaces* (*DSS*), *Non-Dependent Silent Spaces* (*NSS*), the vertical coordinate of the *Center of Ventilation CoV(v)* and the *Functional Lung Spaces* (*FLS*) are shown in the trend graph in percentage values. The values of *NSS* and *CoV(v)* can be visualized considering 0% on the top, those of *DSS* and *FLS* considering 0% on the bottom.

Limitations

This section points at some limitations that need to be taken into account when using EIT in general and SenTec EIT in particular. It does not aim to be a complete list, but rather to raise awareness on the main limitations and contribute to a better data interpretation.

First of all, the output information may be affected by noise in the measurement or by the limitations of the reconstruction model. In order to cancel out most of the errors that may be derived, the current EIT implementation is based, as mentioned earlier, on a time-difference imaging method suited to trace time-varying physiological phenomena. Accordingly, the color intensities in the images displayed are always relative and not based on absolute impedance measurements.

In order to have reliable measurements it is important to wear the belt according to the instructions of use and to pay attention to warnings and contraindications. For example one should avoid using EIT on a patient with pacemaker or with open wounds on the thorax. It is also important to apply enough contact medium for a good signal quality; if this cannot be achieved, the quality of the measurements may be affected. The indication of failing electrodes and signal quality however notifies the user and prompts to take action.

It is worth keeping in mind that significant changes in electrode contact may induce changes in the signal, for example sudden jumps visible in the trend chart of *EELI* and *EILI* (though not affecting their difference, related to the tidal volume). A shift in the signal or other artifacts may also be due to movement or change in patient position, belt reapplication, fluid management or interference with other medical devices[12]. Impedance may be affected by heart or blood movements as well. All these artifacts or signal disturbances can often be excluded or eliminated by adapting the EIT data acquisition parameters, by signal processing or by contextualizing the data acquisition process. Moreover concepts such as *RTS* or *Silent Spaces*, by referring to a breath or a 15 second interval, focus on short term changes that are likely to be unaffected in case of a signal interruption or distortion at a certain time.

Another important point concerns the assumptions about the reconstruction algorithm. Of course, as for any model, there are limitations in the model assumptions and the a priori information. For example EIT is believed to be sensitive to changes in electrical tissue conductivity in a slice with a vertical thickness roughly half the chest width and may not reflect the behavior of the entire lung. Among the assumptions, the models of thorax and lung contours should be considered in the interpretation of results: e.g. in case of lung resection or thoracic deformities, the model may deviate significantly from reality. The breath detection algorithm also relies on certain assumptions, and has influence not only on the indication of respiratory rate, but also on the display of *Stretch Image* and *Silent Spaces*. It is anyway arbitrary how to delimit breaths in certain situations with no regular pattern and there is no consensus on how to define respiratory rate either. Indeed, to allow further analysis on irregular breathing patterns beyond what can be derived from the breath detection, different analysis modes are available, and it is possible to switch between modes at any time.

Finally, another limitation or rather challenge lies in the interpretation of the output information, which may not always be straightforward because changes in impedance and derived indicators between different time points may sometimes hardly be linked to a single possible cause, leading to uncertainty in interpretation. For a better data interpretation it is therefore important to understand the main concepts discussed in this document and have an overview of the different variables that may come into play. Furthermore, the information provided should be considered as an adjunct and evaluated in the context of the clinical situation together with the information provided by other devices.

[12] Frerichs et al. Patient examinations using electrical impedance tomography - sources of interference in the intensive care unit. Physiol. Meas. 32(12), 2011, L1-L10.