

Scientific Paper

Lung divisions for models of cardiopulmonary interaction – preliminary tests

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Abstract

Introduction: The perfusion of a part of the lung depends on its distance from the pulmonary trunk (differences in vascular resistance) and on the horizontal plane (differences in hydrostatic pressure). The aim of this study was to determine the geometric parameters characterising their positions and sizes in order to analyse the diffusion of the ventilation/perfusion ratio.

Material and methods: A developed virtual respiratory system has been supplemented with an appropriate model of pulmonary circulation that uses a lung outline that is divided into parts based on an anatomical atlas and a CT image; it comprises a 3D geometric model of the lungs that was developed using the Inventor CAD software (Autodesk, Inc, San Francisco, USA). Each panel was divided into 2 horizontal and 8 vertical parts; the 16-part division was then modified.

Results: When taking human lungs as a research object and simulating their accompanying physical, biological, or biochemical phenomena, one necessary task is to construct a spatial model of the lungs that takes into account, and maintains awareness of, the limitations of the source of data that is relied upon. The developed modified geometric model of lung division turned out to be useful and was successfully applied to a virtual patient, among others, as part of the VirRespir project.

Conclusions: Finally, we can conclude that the virtual cardiorespiratory system thus elaborated may serve as a proper tool for the preliminary analysis of such complex interactions, considering the elaborated model of the lung's divisions and its future improvements.

Keywords: lung divisions; geometric parameters; modelling; cardiopulmonary interaction.

Introduction

In world medicine, the tendency to treat the circulatory and respiratory systems as a single cardiorespiratory system is growing, due to both the number and types of interactions between these systems and the interrelationships of these systems' pathologies. For example, left-ventricular failure causes pulmonary oedema, which impairs gas exchange. Also, chronic obstructive pulmonary disease, which the WHO ranks third among the top 10 causes of death¹, leads to heart failure, particularly heart attack.^{2,3,15}

The perfusion of a given part of the lungs depends on its distance from the pulmonary trunk (differences in vascular resistance) and from the horizontal plane (differences in hydrostatic pressure). Moreover, ventilation is influenced by gravity due to quasi-hydrostatic pressure and lung compliance

nonlinearity.^{4,6} Therefore, a previously developed virtual respiratory system^{9,12} was supplemented with an appropriate model of pulmonary circulation.¹⁵ Each of the lung lobes was divided horizontally into eight parts and into two vertical planes (front-back, left-right). Each part was characterised by the ratio of its volume to the total volume of the lungs, as well as the coordinates of its centre of mass in a Cartesian coordinate system with the pulmonary trunk as the system origin. Accordingly, this made it possible to determine what proportion of the total lung vasculature would be attributable to a given portion of the lung, the average arterial and venous resistance of the blood-supplying vessels, and the effect of hydrostatic pressure on the transmural pressure in vessels' modification of their resistance. These sets of numbers were analogously used in the lung model.^{9-13,15}

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Authors' contribution:

A – Research concept and design, B – Collection and/or assembly of data, C – Data analysis and interpretation, D – Writing the article, E – Critical revision of the article, F – Final approval of the article.

An important issue in the study of cardiopulmonary interactions is the analysis of the ventilation/perfusion ratio (V/Q).⁴⁻⁶ The V/Q value in different regions of the lungs depends on the position of the body (i.e. the influence of gravity), the distance from the pulmonary trunk ($V/Q > 1$ for distant parts)⁷, and the manner of breathing (e.g. artificial ventilation or natural, spontaneous breathing).

Referring to the works on the flexible modelling of the lungs as a biological object of a spatial nature, two techniques for obtaining source data are inextricably linked. The first is CT technique, as an extension of X-ray technique, which enables one to obtain a solid multilayer model.²⁰ The second is the use of MRI, which allows for the visualisation of the pulmonary tree structure with alveolar endings. Models used in the literature^{18,19,21,23-25} have been made using complex algorithms with a closed structure and are fully integrated with the medical imaging equipment, which often does not allow for interference with the 3D object reconstruction process.

The current work represents a continuation of many years of research on the development of a cardiopulmonary platform, and on the cardiopulmonary system in general, that have been conducted by the Nalecz Institute of Biocybernetics and Biomedical Engineering, Polish Academy of Sciences (IBBE PAS)^{9-13,15-17} in cooperation with the Institute of Clinical Physiology, Italian National Research Council (IFC CNR), Section of Rome, Italy^{8,14}, among others. It can also be applied to the use of computer model simulations when testing new diagnostic methods.¹⁶

In this work, we divided the lungs into 80 parts with the goal of determining the geometric parameters characterising their locations and sizes for V/Q scatter analysis and making simulations more precise and accurate. The aim of this research was to obtain a geometric model of the lungs based on statistically compiled data from Pabst R. et al. anatomical atlas²² and peculiar, often sensitive, data from patient diagnostics. The model thus obtained constituted the input data for the developed method of fragmenting the functional volumes of the lungs in a way that would make it possible to use the obtained subparts of the lung lobes to model the circulatory-respiratory interaction. The purpose of this fragmentation was to develop a procedure that would successfully accomplish the following:

1. The ability to approximate the complex volumes of the pulmonary lobe subparts using a mathematical form that is easy to describe and implement in diagnostic systems;
2. The development of quantitative and qualitative verification of the applied fragmentation and its mathematical approximation.

It should be noted here that while the fragmentation of the lung volume can be approximated quite accurately via CT scan, the division of the lung into lobes was performed using a simplified method; however, this did not affect the final method of lung fragmentation, neither did it change the overall lung volume. It is worth noting, however, that the homogeneous

density of the lungs was assumed while conducting this research, accounting for water density as a reference parameter.

During fragmentation, the individual subparts showed full independence from the others, thus constituting a set of complementary elements without common parts. Constructing such a model allowed the subparts to be approximated, for example, by means of spherical forms.

The replacement model established the same volumetric parameters, but it lost some consistency because the set of substitute subparts was no longer a set of fully complementary elements, and also because it contained common volumes. Because of this, errors may occur in parametric models due to the smaller lung volume that results from accounting for the common volumes, considering that the impact of the interactions of common volumes on the final results of the simulation was unknown at this level. Hence, methods should be proposed for assessing the fragmentation and replacement model that can yield information about the significance of the fragmentation method used for the results of further simulation works.

Moreover, the different regions of the lungs function differently due to the influence of gravity and their distances from the pulmonary trunk. Taking into account the ventilation aspects, the upper (independent) regions of the lungs are expanded, and their lower (dependent) parts are compressed due to their weight because the independent regions lying on the dependent ones compress them, and the dependent regions are under a certain amount of strain in relation to the independent ones that surround them. Since lung compliance is non-linear (the more the lung is dilated, the less compliance), the dependent areas may be more ventilated than the independent areas. Given the fact that regional bronchial occlusion occurs with small volumes, the dependent areas that are most compressed may not be ventilated at all. In pulmonary circulation, the effects of gravity are related to hydrostatic pressure, which dilates the vessels in the dependent areas (positive hydrostatic pressure) and causes the vessels in the independent areas to narrow or collapse (negative pressure). In addition, the distance between the pulmonary trunk and a given area affects vascular resistance and thus blood flow. Moreover, heterogeneous ventilation may affect regional perfusion due to hypoxia caused by pulmonary vasoconstriction. In order to account for all these aspects in the simulations, each lobe of the lungs was divided into 16 different parts.

Materials and Methods

Using a lung outline divided into parts based on the atlas of human anatomy²² and a CT image, a 3D geometric model of the lungs was developed using the Inventor CAD software (Autodesk). Each panel was divided into two horizontal and eight vertical planes; this 16-part division was then modified.

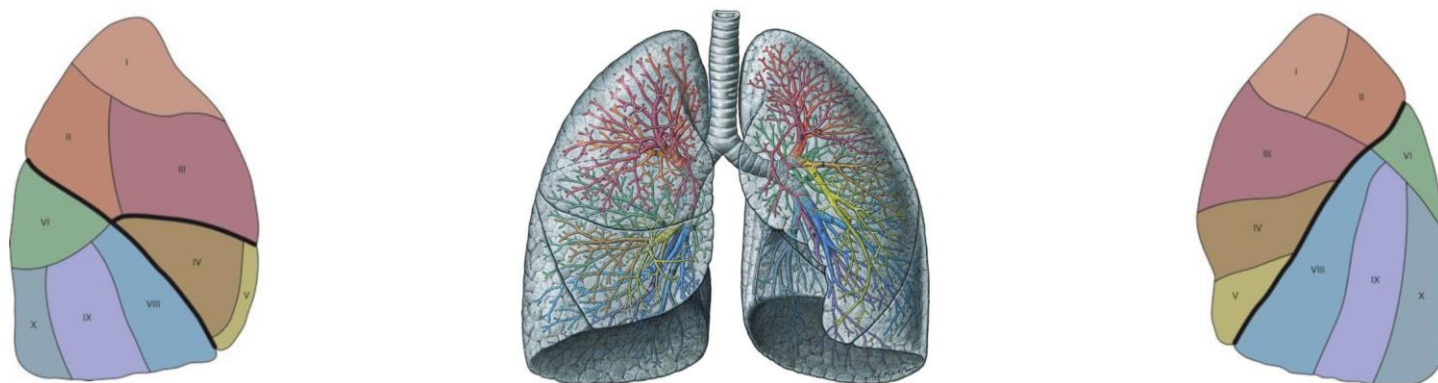


Figure 1. Source views of human lungs including the division of lungs into functional lobes²²

One of the many tasks related to the analysis of human lungs as research objects and the simulation of the associated physical, biological, and biochemical phenomena is the preparation of a spatial model of the lungs that takes into account, and maintains awareness of, the limitations of the source data used.

Practices that had been widely used in industrial spatial modelling were adopted to develop a method for modelling lungs, the procedure for which includes the following five steps:

- Obtaining the general geometry of the left and right lungs using available source data;
- Dividing the general geometry of the right and left lungs into the functional parts of the lung – the so-called pulmonary lobes – using available source data;
- Locating the reference point for the spatial model, the so-called pulmonary trunk, in the obtained lung volume;
- Fragmenting the pulmonary lobes using the Cartesian frame of reference as a discriminant;
- Formulating the matrix describing the features of the lungs with the pulmonary trunk as a reference point for the spatial features of the obtained lung model.

Basic method

This approach relied on the lung views provided in the anatomical atlases shown above (**Figure 1**) to derive the lung envelope in the baseline views obtainable by standard optical methods. The atlases used, however, lack information on the scaling of the models and their spatial relationships to one another, and therefore could not be considered sufficient for spatial lung modelling in this particular application. While these resources can serve as a starting point, allowing for the specification of specific requirements for source data obtained with the use of other cognitive methods, necessary to build a lung model with a minimum set of parameters necessary for the construction of future mathematical models.

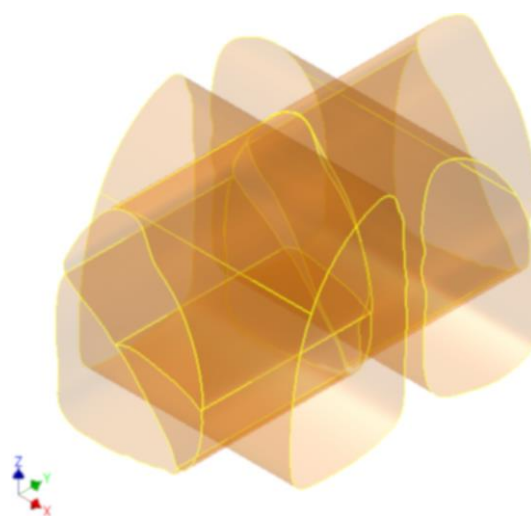


Figure 2. Visualisation of the free surfaces that limit the volumes of the left and right lungs, including their divisions into functional parts using side views.

Simplified hybrid method

The hybrid method allowed the data that was used in the basic method to be supplemented with data derived indirectly from medical imaging, having its source in x-ray imaging. It was possible to supplement the basic model with several variable cross-sections, giving the lung model several special features, such as a simplified heart cavity (resulting from the location of the heart muscle between the left and right lung and the spine), muscles, and other tissues directly penetrating the lung volume obtained by the basic method. This method used for the spatial modelling of the lungs, which employed the basic method outlined above and the horizontal sections made using the CT-imaging technique, was considered by the authors of the atlas of human anatomy²² to result in reliable, statistically correct representations of human anatomy (**Figure 3**).

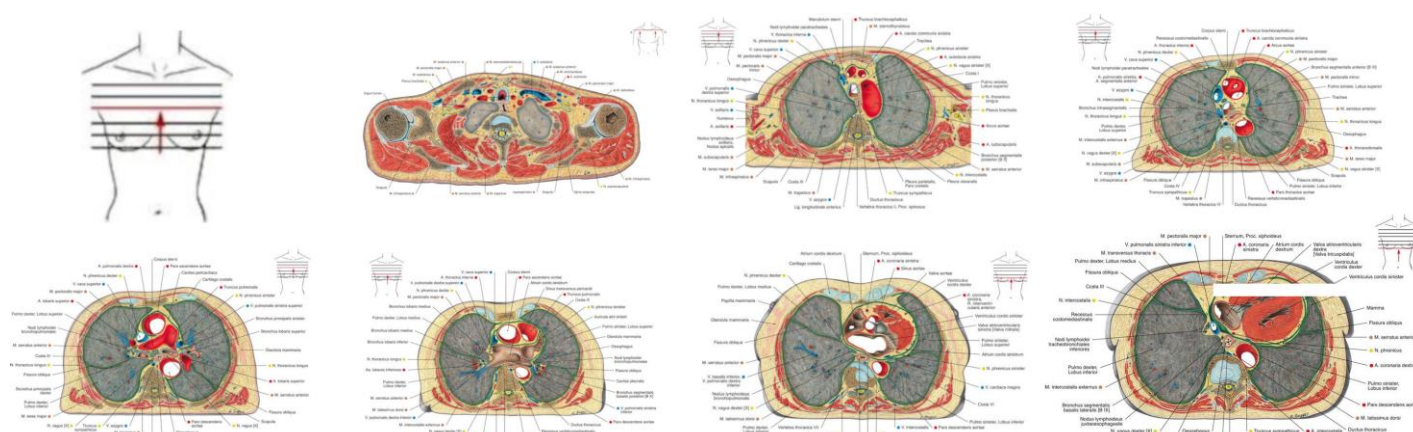


Figure 3. Source images of body sections with emphasis on the lungs²²

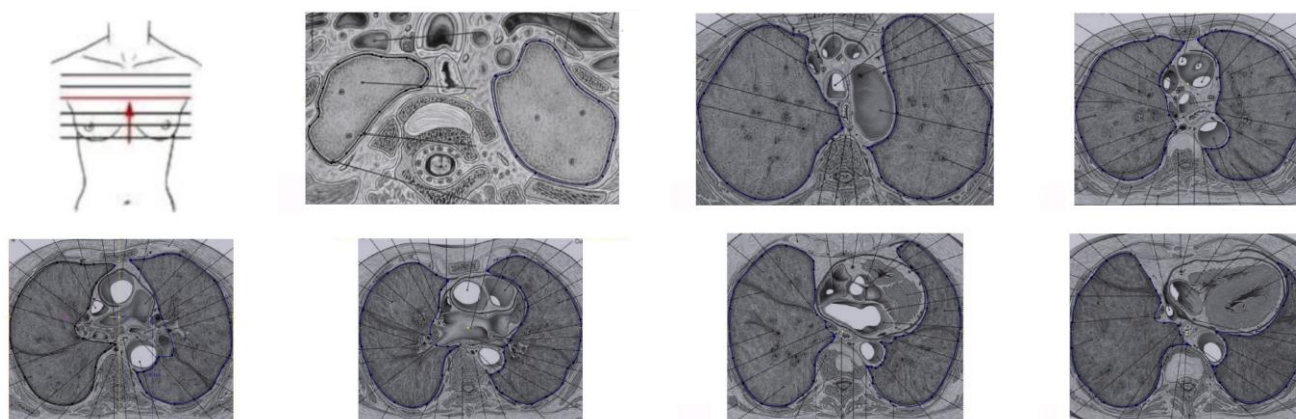


Figure 4. Visualisation of the lung envelope in the contours²², determined by vector tracing using the minimum energy method

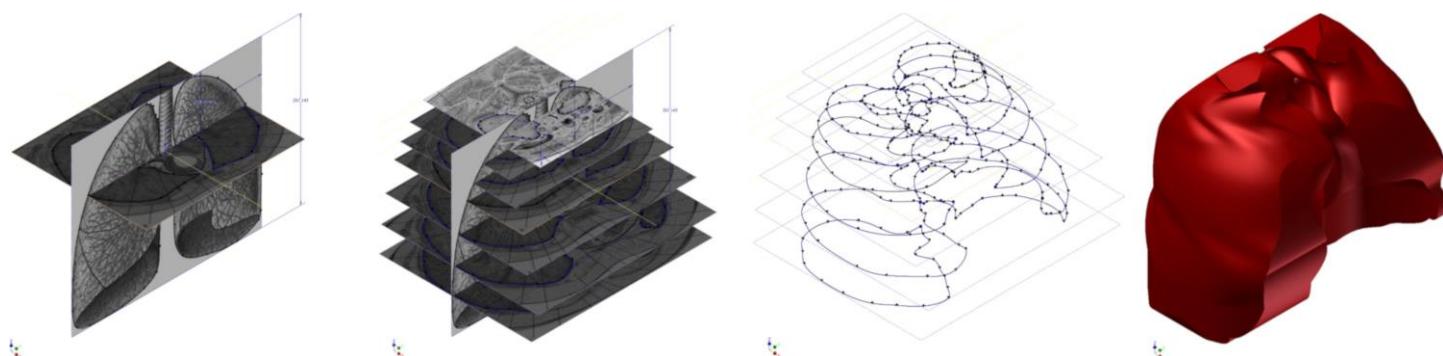


Figure 5. Methodology for assembling contour lines/vector envelopes to create a perpendicular/long-axis extrusion of a variable-section body based on the above-mentioned envelopes

In addition, Pabst R. et al. anatomical atlas²² presents lung cross-sections that have not necessarily come from the same person; the integrity of the data used is thus debatable, although the assumption here is to recognise the images contained in Pabst R. et al. anatomical atlas²² as verified by human anatomy experts in terms of compliance with the statistical model of a human.

Another limitation of this data is the lack of extreme sections that would enable proper completion of extension in the apical

and lower parts of the lungs, which may result in significant errors in modelling these zones.

Moreover, the location of the cross-sections along the long axis has been shown visually in the general view. The distance between the individual cross-sections is not known, which necessitates a certain amount of interpolation vis-à-vis the position of the individual cross-sections used to build the model. In addition, there is no data on the size or height of the statistical human model, which forces the tendency to obtain

a model in a specific scale. In this case, one of the x-ray images was used as a model to scale the height of the lung model. The obtained height was determined to be ~257 mm and corresponds to the scaling shown in **Figure 6**. All the drawings from the atlas of human anatomy²² with the vector of the long axis of the body as one of the basic dimensions were scaled in this way, and the second basic dimension resulted from maintaining a constant scale factor in both perpendicular directions when scaling the raster image.

There was also no data related to the unequivocal location of the so-called pulmonary trunk (**Figure 7**), which was a reference for the construction of a mathematical model of the gas exchange phenomenon that was determined by an expert in human anatomy in relation to the planar view of the lungs which, unfortunately, was not fully visualised using the available cross-sections presented in the atlas of human anatomy.²²

One of the greatest simplifications was the division of the lungs into lobes due to the rather simplified way of showing the divisions of the right and left lungs into lobes using lateral views of the lungs and establishing the boundaries as perpendicular rather than free surfaces that depended in any way on the location of the deep view/section.

Although the digital modelling of the lungs and the spatial interpolation of the reference points were completed with the Inventor CAD software (Autodesk, Inc, San Francisco, USA), the presented method is quite universal and can be adapted to any 3D-modelling software environment.

The source materials were prepared using open-source software such as GIMP (GNU Image Manipulation Program), for converting the raster graphics from scans in the atlas of human anatomy²² to an Inkscape-compatible format, and Inkscape (Inkscape Team, GPL) having the Better DXF Output script that was used for vectorising and optimising the Bezier curves and exporting the vector envelopes in DXF format.

Results

The original representation of the lung model, visualised as a solid model after division into lobes, is presented in **Figure 8**.

Figure 9 shows the universal representation of the lung model, with the visualisation of the alveolar model divided into lobes.

The model illustrated in **Figure 8** is a solid lung model made from the anatomical data contained in the atlas of human anatomy²², supplemented by critical data on scaling the 3D object. The object imaging human lungs was separated into independent zones representing the lobes in terms of basic functional division. The colours used in the solid model represent the volumes assigned to the individual lobes.

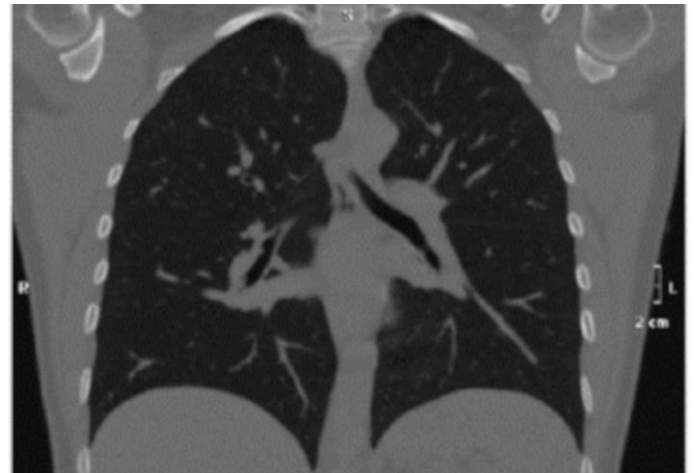


Figure 6. X-ray image used as a model to scale the height of the lung model

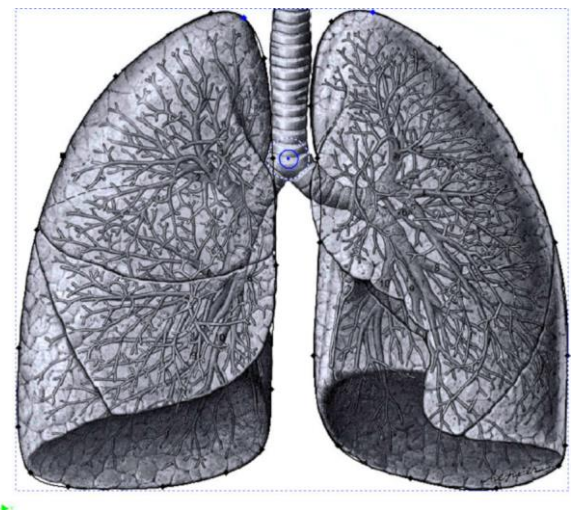


Figure 7. Location of the reference position of the pulmonary trunk made by marking the centre of the circle inscribed in the pulmonary canals

The model illustrated in **Figure 9** is a bubble representation of the solid model from **Figure 8**, presented in order to analyse the correctness and validity of the assumptions that were made, as well as to verify the fragmentation methodology of the presented biological model. The bubble model, due to its universal nature, did not transfer all the features and spatial relations included in the solid model. However, the distinguishing features of the defragmented solid model – as a rule – in terms of volume and the mutual position of the centre of gravity of the fragments and sets 01, 02, 03, 04, 05, 12, 35, and 15 were in accordance with the solid model before defragmentation. The colours used in the bubble model represent the volumes assigned to the individual lobes, and the bubbles to the individual fragments.

The elaborated model of lung divisions thus modified was used in the virtual patient developed at the Institute, which was initiated by the VirRespir project.⁵

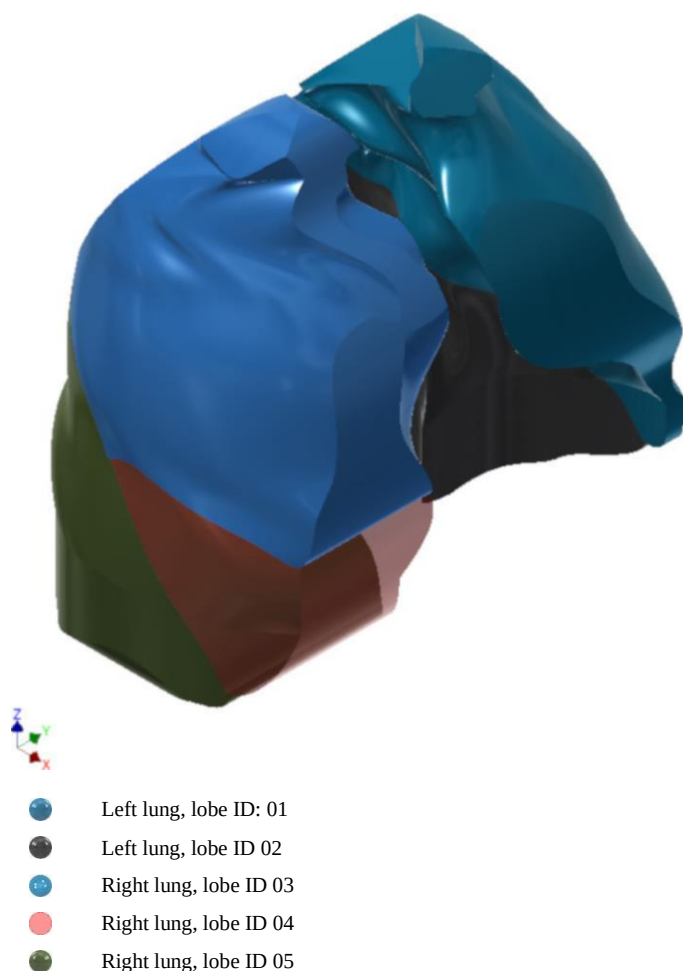


Figure 8. Original representation of the lung model, visualization of the solid model divided into lobes

Discussion

The spatial modelling of human lungs resulted in a preliminary model that used data that is available in the literature to be used as source data for building simple 3D models of biological objects such as human organs.¹⁸⁻²¹ In such cases, spatial modelling always encounters limitations related to the lack of certain data considered vital to certain modelling steps. Thanks to either expert knowledge or the use of certain simplifying assumptions, we can attempt to interpolate, scale, or adjust some parameters, thus allowing the integrity and consistency of models to be maintained while remaining cognizant of their imperfections.

The creation of a digital model of a biological object²³⁻³³, supplemented by objects with specific volumes and spatial relations, allowed only a limited approximation of its essence due to the assumptions and simplifications that had to be made. Considering recently how important the use of modelling in the diagnosis of lung cancer^{26,31,33} or during the Covid-19 pandemic has become.²⁹

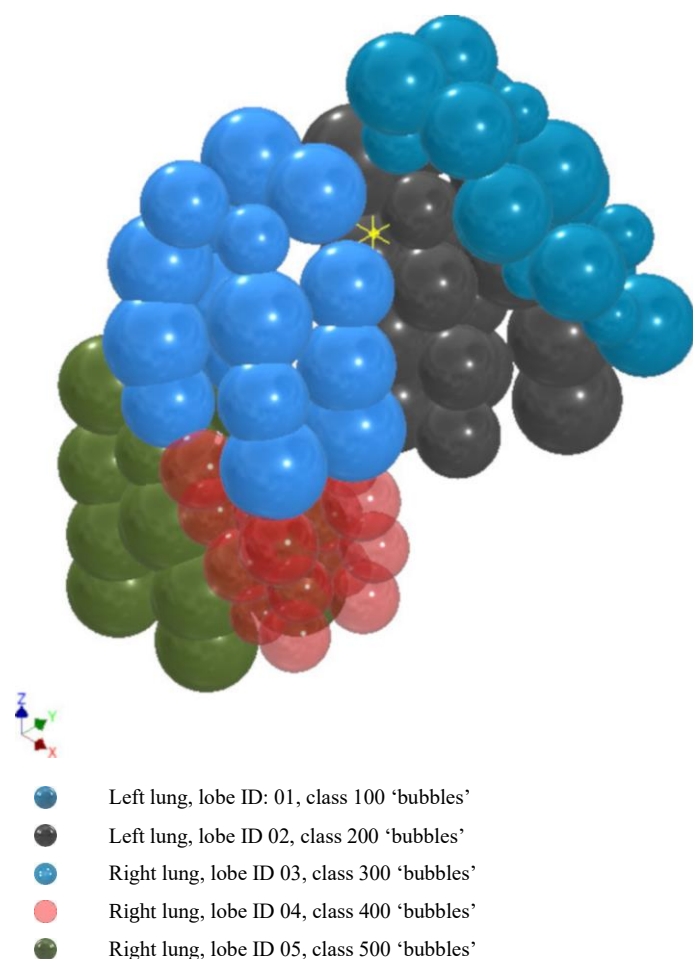
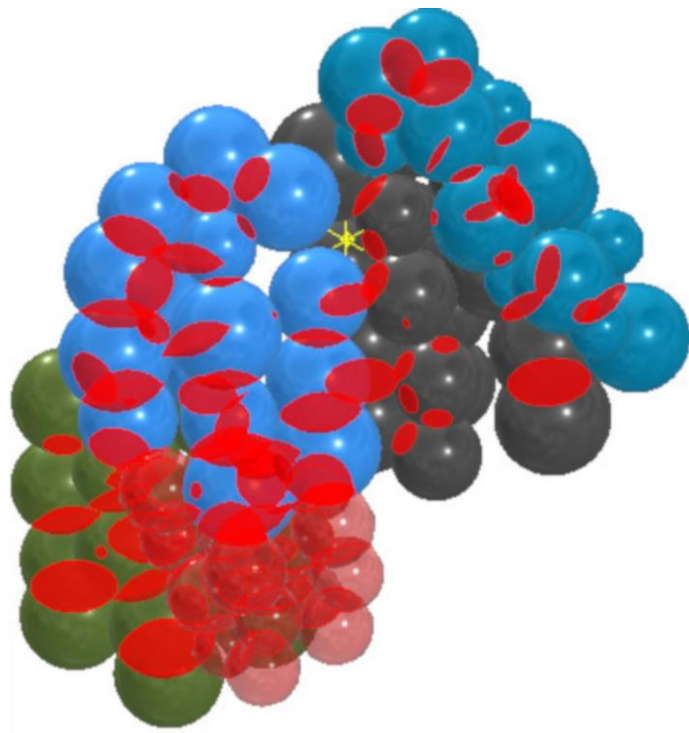


Figure 9. Universal representation of the lung model with the visualisation of the alveolar model divided into lobes

After the spatial model was divided into five basic lobes, these lobes were further fragmented into 16 parts. Each of these parts is described by the location of the centre of the selected element in relation to the reference point determined and referred to as the centre of the so-called pulmonary trunk and its volume (see Appendix). In order to be able to compare the models created by different techniques and the fragmentation and selection of different parameters, it is possible to present such a matrix in a universal way using simple geometric solids. In the current work, spheres were used to represent the pulmonary alveoli. However, it should be remembered that, in the case of the fragmentation of the lung model, the volumes obtained directly did not have common spaces; therefore, such phenomena as so-called volume collisions – in the case of such universal models – may occur due to subsequent simplifications imposed on the model. The distinguishing feature of such models may be the percentage rates of volumetric collisions in the areas of specific lobes, the left lung, the right lung, and the global collision rate (**Figure 10**).



ID = 15
WKEP₁₅ = 100/080
VK₁₅ = 2.11398403E+05
WKEV₁₅ = 0.073873
WKEV₁₅ = 7.39%

V₁₅ = 2861641.982
P₁₅ = 409963.734
oX₁₅ = -14.033
oY₁₅ = -9.257
oZ₁₅ = -66.475

Figure 10. Visualisation and parameterisation of the collision phenomenon in the alveolar lung model, set ID: 15

For the universal/comparative model, both qualitative and quantitative indicators allowing for the comparison of individual lung models were proposed and determined.

The first indicator is the WKGN_{ID/ID*} group collision index in terms of the total number of collisions, regardless of its spatial extent, which describes the number of volumetric collisions in relation to the number of objects included in the set. For the above model, the set consisted of 80 objects/bubbles, and the number of counted mutual volumetric collisions was 100. The first indicator can be calculated using this equation:

$$WKGN_{ID/ID*} = LK_{ID}/LO_{ID*} \quad \text{Eq. 1}$$

where:

LK_{ID}: the number of volume collisions counted

LE_{ID*}: the number of elements in the set

WKGN_{ID/15} = LK_{ID}/LO₁₅; ID, ID*: 01, 02, 03, 04, 05, 12, 35, 15

WKGN_{ID1/12} = LK_{ID1}/LO₁₂; ID1: 01, 02, 12

WKGN_{ID2/15} = LK_{ID2}/LO₃₅; ID2: 03, 04, 05, 35

It is worth noting that the number of collisions is not unlimited, although the number of possible collisions is much smaller than theorised/originally assumed. As this parameter indicates a certain number of collisions rather than the quality of fragmentation, an additional indicator has been proposed.

The second indicator is the WKGV_{ID/ID*}, or the group collision index, which qualitatively describes the independence of the alveolar model representation, thus providing information about the correctness of the method and the degree of the lung model's fragmentation. The second indicator is calculated using the following equation:

$$WKGV_{ID/ID*} = VK_{ID}/VC_{ID*} \quad \text{Eq. 2}$$

where:

VK_{ID}: collision cumulative volume

VC_{ID*}: ID* set volume

WKGV_{ID/15} = VK_{ID}/VC₁₅; ID, ID*: 01, 02, 03, 04, 05, 12, 35, 15

WKGV_{ID1/12} = VK_{ID1}/VC₁₂; ID1: 01, 02, 12

WKGV_{ID2/35} = VK_{ID2}/VC₃₅; ID2: 03, 04, 05, 35

In the case of our model, this indicator amounted to about 7.39% of collisions in the entire model's volume. While this index describes the fragmentation of the model in a general way, it describes in some detail the impact of fragmentation on the individual lobes. As a high degree of inconsistency in a specific area may, but does not have to, significantly affect the assessment of the usefulness of the applied fragmentation, two additional indicators were proposed: the share and significance of the impact on the spatial model.

One of them, the WUOV_{ID/ID*}, is an index of the share of the object's volume in the set volume with a higher degree of organisation, i.e. with ID: 01 or 02 in the set with ID: 12 or 15; or with ID: 03, 02, or 05 in the set with ID: 35 or 15; or with ID: 12 or 35 in the file with ID: 15.

The WUOV_{ID/ID*} index can be obtained by this equation:

$$WUOV_{ID/ID*} = VC_{ID} / VC_{ID*} \quad \text{Eq. 3}$$

where:

VC_{ID}: the ID set volume

VC_{ID*}: the ID* set volume

WUOV_{ID/15} = VK_{ID}/VC₁₅; ID, ID*: 01, 02, 03, 04, 05, 12, 35, 15

WUOV_{ID1/12} = VK_{ID1}/VC₁₂; ID1: 01, 02

WUOV_{ID2/35} = VK_{ID2}/VC₃₅; ID2: 03, 04, 05

The second is the WIW_{VID/ID*}, or the indicator of the significance of the impact of the collisions occurring in the facility on the facilities with a higher degree of organisation (see guidelines for the indicator WUOV_{ID/ID*}). The indicator WUOV_{ID/ID*} can be calculated by this equation:

$$WIW_{VID/ID*} = WKGV_{VID/ID*} / WUOV_{VID/ID*} \quad \text{Eq. 4}$$

where:

WKGV_{VID/ID*}: group collision indicator

WUOV_{VID/ID*}: volume share indicator

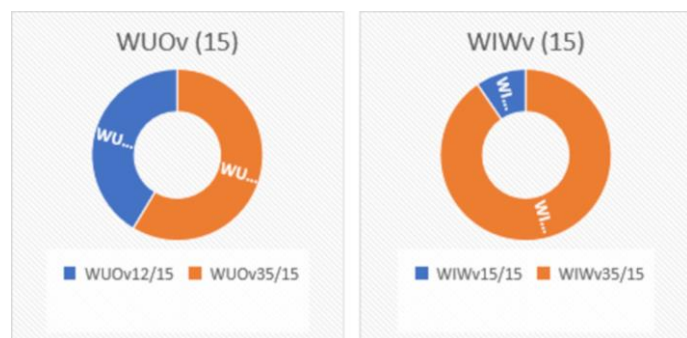
WIW_{VID/15} = WKGV_{VID/15} / WPO_{VID/15}; ID, ID*: 01, 02, 03, 04, 05, 12, 35, 15

WIW_{VID1/12} = WKGV_{VID1/12} / WPO_{VID1/12}; ID1: 01, 02

WIW_{VID2/35} = WKGV_{VID2/35} / WPO_{VID2/35}; ID2: 03, 04, 05

Table 1. Selected quantitative and qualitative parameters for the set ID: 15 and its subsets ID: 12, 35

Sets	Quantitative indicators	Qualitative indicators
ID = 15	WKG _{n15/15} = 100/080 WKG _{v15/15} = 0.073873	WUO _{v15/15} = 1 WIW _{v15/15} = 0.073873
ID = 12	WKG _{n12/12} = 030/032 WKG _{v12/12} = 0.023283 WKG _{v12/15} = 0.005422	WUO _{v12/15} = 0.413985 WIW _{v12/15} = 0.002245
ID = 35	WKG _{n35/35} = 051/048 WKG _{v35/35} = 0.109611 WKG _{v35/15} = 0.036132	WUO _{v35/15} = 0.586015 WIW _{v35/15} = 0.021174

**Figure 11.** Donut charts representing the share and quantitative impact of individual sets ID: 12, 35 on the structural elements of the higher order ID: 15

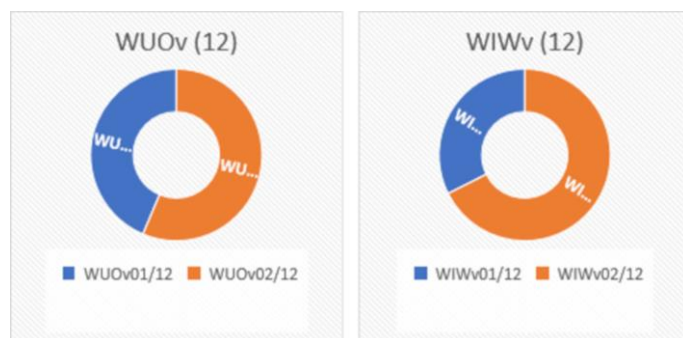
Considering the lungs as exclusive sets with ID: 12 and 35, quantitative and qualitative indices were determined for the entire lung model. As we can see, despite the comparable contribution to lung volume formation in the WUOv (15) diagram, the collision errors for the set with ID: 35 significantly affected the quality of the entire lung model (see the WIWv (15) diagram, **Table 1** and **Figure 11**).

For the left lung, the ID: 12 quantitative indicator WKG_{nID/ID*} for the sets with ID/ID*: 12, 01, 02 did not exceed 1, which indicates a fairly high probability of high consistency for this lung model. Indeed, this lung has a qualitative index several times lower WKG_{vID/ID} for the sets with ID/ID*: 12, 01, and 02 compared to the global index WKG_{n15/15}. Qualitative indicators, both during the creation of the set with ID: 12 and as a component of the set with ID: 15, behaved stably – as expected (see **Table 2**, **Figure 12**).

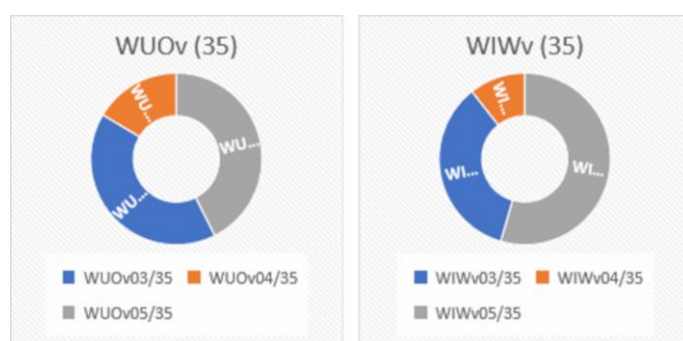
For the right lung, the ID: 35 quantitative indicator WKG_{nID/ID*} for the sets with ID: 35, 03, 04 exceeded 1, which indicates a fairly high probability of low consistency. Indeed, this lung has a higher qualitative index WKG_{vID/ID*} for ID/ID*: 35, 04, and 05 than the global index WKG_{n15/15}. Qualitative indicators, both when creating the set with ID: 35 and as a component of the set with ID: 15, behaved unexpectedly for ID: 35, 03, and 05. The amount of influence of the collision errors on WIW_{v35/15} caused this index to reach a high value of 10E-2; this was due to the collision errors that appeared in the objects with ID: 03 and 05 (especially ID: 05). There was a high impact on both higher-order structural objects with ID: 35 and 15 (see graph WIWv (35), **Table 3**, **Figure 13**).

Table 2. Selected quantitative and qualitative parameters for the set ID: 12 and its subsets ID: 01, 02

Sets	Quantitative indicators	Qualitative indicators
ID = 12	WKG _{n12/12} = 030/032 WKG _{v12/12} = 2.33%	WUO _{v12/15} = 0.413985 WIW _{v12/15} = 0.002245
ID = 01	WKG _{n01/01} = 011/016 WKG _{v01/12} = 0.004779	WUO _{v01/12} = 0.436400 WIW _{v01/12} = 0.002085
ID = 02	WKG _{n02/02} = 016/016 WKG _{v02/12} = 0.007715	WUO _{v02/12} = 0.563600 WIW _{v02/12} = 0.004348

**Figure 12.** Donut charts representing the share and quantitative impact of individual sets ID: 01, 02 on the structural elements of the higher order ID: 12**Table 3.** Selected quantitative and qualitative parameters for the set ID: 35 and its subsets ID: 03, 04, and 05

Sets	Quantitative indicators	Qualitative indicators
ID = 35	WKG _{n35/35} = 051/048 WKG _{v35/15} = 0.036132	WUO _{v35/15} = 0.586015 WIW _{v35/15} = 0.021174
ID = 03	WKG _{n03/03} = 019/016 WKG _{v03/35} = 0.01595	WUO _{v03/35} = 0.410033 WIW _{v03/35} = 0.006540
ID = 04	WKG _{n04/04} = 018/016 WKG _{v04/35} = 0.012111	WUO _{v04/35} = 0.164328 WIW _{v04/35} = 0.001990
ID = 05	WKG _{n05/05} = 014/016 WKG _{v05/35} = 0.024007	WUO _{v05/35} = 0.425639 WIW _{v05/35} = 0.010218

**Figure 13.** Donut charts representing the share and quantitative impact of individual sets ID: 03, 04, and 05 on the structural elements of the higher order ID: 35

Conclusions

The lung division developed for this study was used in virtual cardiorespiratory system models. This work is the result of research on the cardiovascular and respiratory systems carried out at the Institute. The next step will be to use the obtained data to numerically simulate a model of cardiopulmonary interaction

and then verify it using a physical model. This model was both useful and convenient for the preliminary testing of the modelling of cardiorespiratory systems, but it needs to be refined in order to obtain a suitable resolution.

Analysis of its qualitative parameters uncovered the need to improve the lung model's fragmentation – specifically the two largest functional lobes of the right lung. The solution may be either to change the assumptions regarding the technique of solid fragmentation or to increase the resolution of the fragmentation. Both courses of action would not be mutually exclusive, and both may even improve the legitimacy of the research that is intended to take place in the near future related to the above-mentioned issues.

The methodological differences in creating different lung models stem directly from the adopted limit assumptions and the permissible simplifications required to obtain an understandable and consistent mathematical model. In the case of the work carried out for the purposes of the VirRespir project⁵, it was assumed that a spatial lung model should reflect the significance of the influence of the direction of the gravity vector on the cardiorespiratory model – hence the division based on subparts' centres of gravity. As for the work on the alveolar structure, which was based on bifurcation algorithms and MRI imaging, the emerging model was so complicated that the total lung volume necessary for a correct mathematical description

would be difficult to obtain with a low enough level of detail (generation) for the VirRespir project.⁵

It seems that an elaborated virtual cardiorespiratory system may be the proper tool for preliminary analysis of complex interactions in both spontaneous breathing and artificial ventilation conditions, considering the elaborated model of lung divisions. This work will be continued using the CT images in the near future.

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Conflicts of interest

Authors of the manuscript “Lung divisions for models of cardiopulmonary interaction – preliminary tests” state that there are no Conflicts of Interest to declare.

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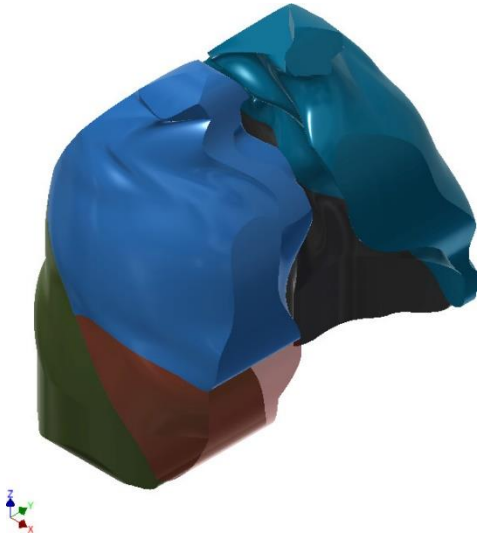
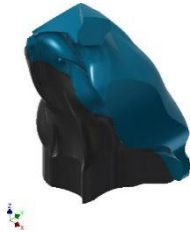
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Appendix A. Supplementary data

Table A1. Original representation of the lung model, visualization of the solid model divided into lobes

Lungs, ID = 15, solid model visualization, isometric view	
ID = 15 	ID = 12  ID = 35 
Left lung, lobe ID 01 Left lung, lobe ID 02 Right lung, lobe ID 03 Right lung, lobe ID 04 Right lung, lobe ID 05 ID01, 02, 03, 04, 05 – functional pulmonary lobes; 12, 35 – left and right lung, respectively; 15 – lungs as a whole	

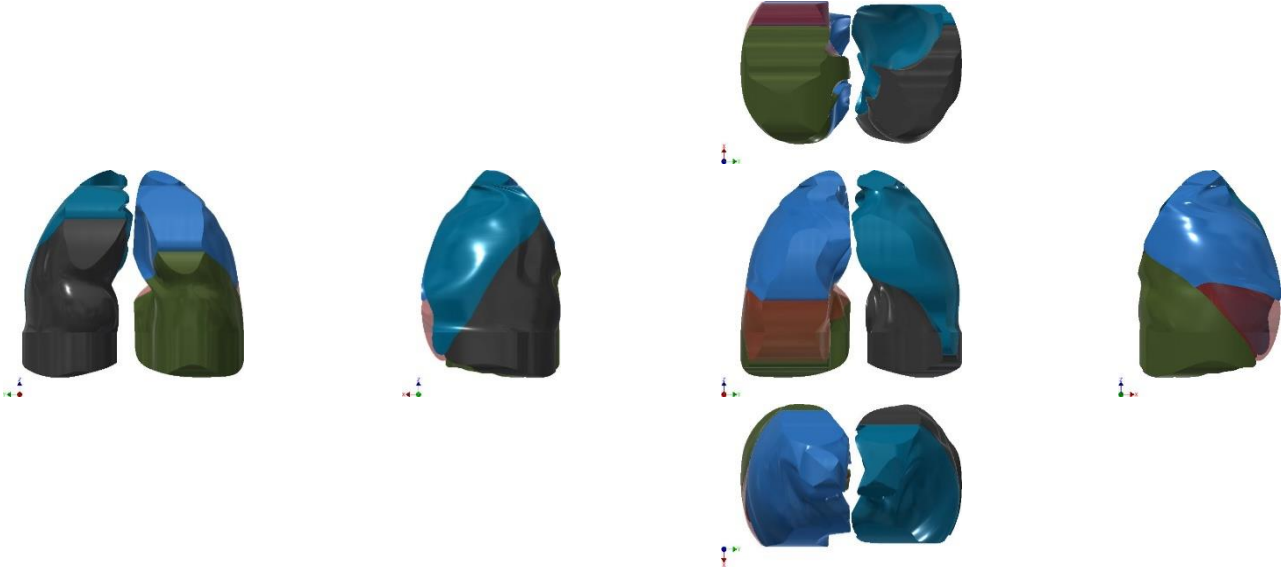
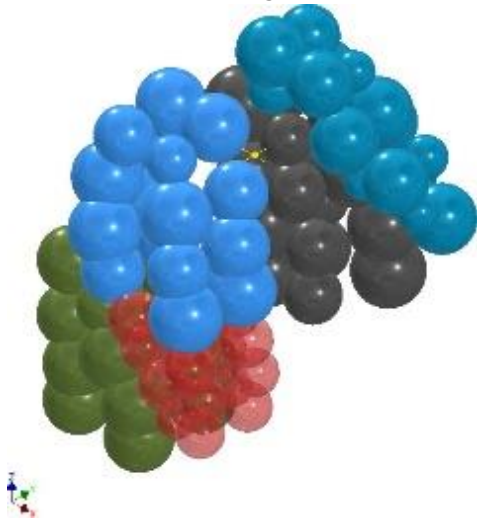
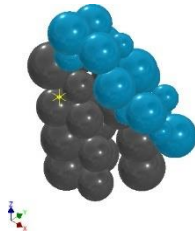
Lungs, ID = 15, solid model visualization, classic views	
	

Table A2. Universal representation of the lung model, source matrix of the alveolar model broken down into bubbles

Lungs, ID = 15, description of alveolar features broken down into lobes with ID: 01, 02, 03, 04, 04																	
01	B _{ID}	oX	oY	oZ	V	02	B _{ID}	oX	oY	oZ	V	03	B _{ID}	oX	oY	oZ	V
	101	12,60	46,11	38,34	75522		201	-35,84	64,14	-37,53	38473		301	3,03	-38,42	42,20	81543
	102	-34,29	49,48	37,94	87331		202	-68,36	57,35	-23,47	114345		302	-37,04	-37,24	40,38	85751
	103	7,24	81,62	27,23	18850		203	-34,57	107,51	-38,71	48357		303	-0,47	-73,73	32,88	32328
	104	-35,49	83,85	32,61	60861		204	-65,56	97,58	-27,17	57920		304	-39,11	-76,91	34,39	56680
	105	30,41	50,01	2,00	72124		205	-29,70	65,93	-75,89	60047		305	25,38	-41,47	4,68	66047
	106	-37,86	54,47	6,13	35789		206	-66,73	62,54	-77,06	66820		306	-47,96	-40,70	4,45	66851
	107	20,40	90,10	-1,33	86261		207	-22,56	110,87	-77,13	113189		307	14,00	-83,81	2,22	94728
	108	-32,11	93,62	1,46	82042		208	-65,89	100,51	-76,45	54601		308	-46,13	-84,66	3,19	93445
	109	50,72	74,68	-32,62	75527		209	-18,35	71,15	-116,14	50129		309	38,12	-54,63	-24,09	76280
	110	-2,02	81,14	-31,35	74101		210	-59,56	66,79	-114,70	109175		310	-41,94	-54,59	-23,58	99983
	111	40,14	112,35	-37,50	22791		211	-4,96	120,44	-116,95	99708		311	30,46	-98,18	-25,04	61957
	112	-0,37	118,23	-33,55	59062		212	-53,79	110,26	-115,78	55641		312	-33,44	-100,16	-24,25	96243
	113	56,03	90,25	-61,20	17910		213	-18,28	70,36	-151,90	50986		313	43,69	-56,33	-54,20	94927
	114	6,43	90,48	-61,92	14534		214	-58,57	69,24	-152,44	98519		314	-24,52	-55,46	-48,19	53002
	115	45,20	121,89	-87,15	87561		215	-2,56	121,90	-150,65	106376		315	40,04	-101,20	-54,17	98433
	116	12,96	122,76	-73,50	48832		216	-53,52	111,22	-154,14	62707		316	-21,75	-104,44	-48,32	64221
04	B _{ID}	oX	oY	oZ	V	05	B _{ID}	oX	oY	oZ	V						
	401	42,00	-57,86	-73,88	20959		501	-26,92	-53,16	-75,85	39039						
	402	2,37	-52,92	-69,99	46491		502	-66,45	-59,96	-60,47	94998						
	403	41,23	-108,01	-73,86	22337		503	-27,60	-107,02	-75,85	40376						
	404	2,89	-109,32	-69,83	43798		504	-64,52	-98,24	-63,67	81265						
	405	50,70	-58,45	-88,29	41015		505	-13,46	-49,46	-104,38	99817						
	406	10,44	-52,57	-87,24	26424		506	-63,54	-60,07	-103,26	58346						
	407	48,73	-107,39	-88,31	43227		507	-14,25	-109,94	-104,62	101250						
	408	11,48	-110,75	-87,45	26118		508	-63,00	-102,26	-103,15	64928						
	409	60,71	-60,39	-106,83	28902		509	4,24	-51,35	-132,61	90918						
	410	30,31	-59,55	-106,16	32545		510	-53,00	-59,49	-132,05	82933						
	411	59,98	-105,61	-106,92	25438		511	6,53	-112,08	-132,71	88072						
	412	30,12	-111,48	-106,19	35937		512	-50,76	-106,95	-132,06	81142						
	413	56,78	-63,11	-133,88	31411		513	7,27	-54,51	-156,57	84829						
	414	39,71	-59,20	-128,96	17289		514	-52,63	-60,39	-159,01	85122						
	415	56,48	-104,92	-134,05	30314		515	8,70	-111,28	-156,92	88038						
	416	39,55	-110,00	-128,83	17701		516	-50,38	-106,51	-159,70	87873						
B _{ID}	bubble identification tag																
oX	position of the bubble center in the X axis with respect to the pulmonary trunk (mm)																
oY	position of the bubble center in the Y axis with respect to the pulmonary trunk (mm)																
oZ	position of the bubble center in the Z axis with respect to the pulmonary trunk (mm)																
V	bubble volume (mm ³)																

Table A3. Universal representation of the lung model, visualization of the alveolar model divided into lobes

Lungs, ID = 15, vesicular model visualization, isometric view	
ID = 15 	ID = 12  ID = 35 
Left lung, lobe ID 01, class 100 "bubbles" Left lung, lobe ID 02, class 200 "bubbles" Right lung, lobe ID 03, class 300 "bubbles" Right lung, lobe ID 04, class 400 "bubbles" Right lung, lobe ID 05, class 500 "bubbles" ID01, 02, 03, 04, 05 – functional lobes; 12, 35 – left lung, right lung; 15 – whole lungs	

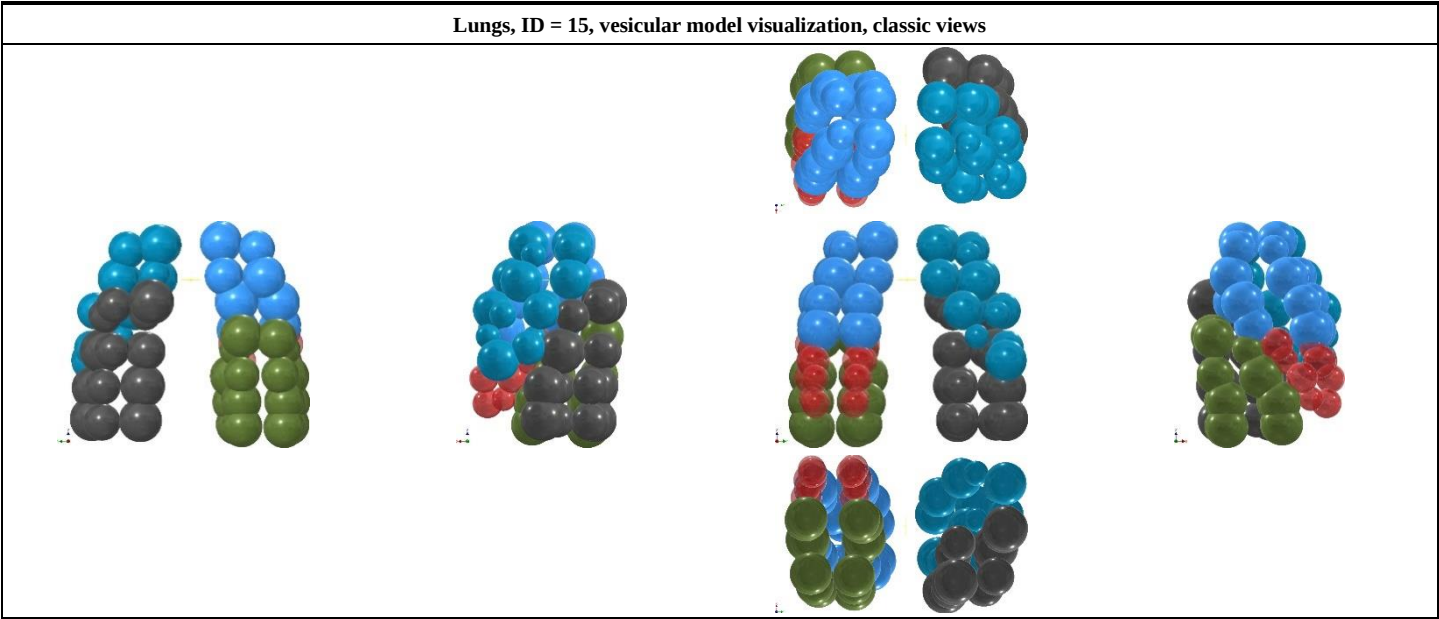
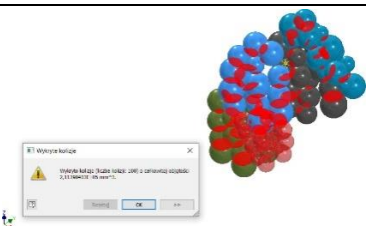


Table A4. Universal representation of the lung model, visualization of the alveolar model taking into account the phenomenon of collision
(continued on the next page)

Lungs, ID = 15, visualization and parameterization of the collision phenomenon		
	ID = 15 WKEp ₁₅ = 100/080 VK ₁₅ = 2.11398403E+05 WKEv ₁₅ = 0.073873 WKEv ₁₅ = 7.39%	V ₁₅ = 2861641.982 P ₁₅ = 409963.734 oX ₁₅ = -14.033 oY ₁₅ = -9.257 oZ ₁₅ = -66.475
V _{ID}	the volume of the object [mm ³]	
P _{ID}	area of the object given in [mm ²]	
oX _{ID}	the position of the COG center of the object relative to the reference point [mm]	
WKEp _{ID}	a quantitative index of collisions per number of set elements [ul]	
VK _{ID}	collision volume [mm ³]	
WKEv _{ID}	collision quantification in relation to the object volume [ul, %]	
ID	01, 02, 03, 04, 05 – functional lobes of the lungs; 12, 35 – left lung, right lung; 15 – whole lungs	

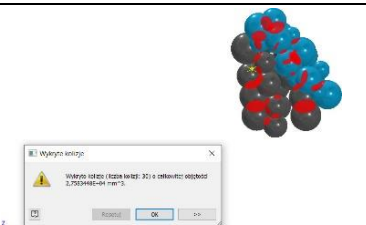
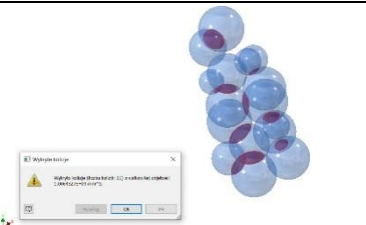
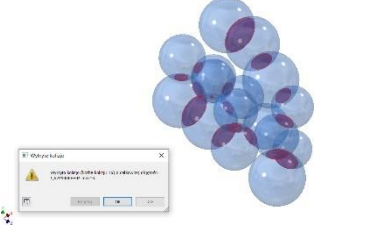
Left lung, ID = 12, visualization and parametrization of the collision phenomenon		
	ID = 12 WKEp ₁₂ = 030/032 VK ₁₂ = 2.7583448E+04 WKEv ₁₂ = 0.023283 WKEv ₁₂ = 2.33%	V ₁₂ = 1184676.326 P ₁₂ = 168032.749 oX ₁₂ = -20.008 oY ₁₂ = 86.251 oZ ₁₂ = -60.229
	ID = 01 WKEp ₀₁ = 011/016 VK ₀₁ = 1.0064327E+04 WKEv ₀₁ = 0.019467 WKEv ₀₁ = 1.95%	V ₀₁ = 516992.849 P ₀₁ = 76217.892 oX ₀₁ = 6.702 oY ₀₁ = 82.961 oZ ₀₁ = -12.800
	ID = 02 WKEp ₀₂ = 016/016 VK ₀₂ = 1.6249000E+04 WKEv ₀₂ = 0.024336 WKEv ₀₂ = 2.43%	V ₀₂ = 667683.478 P ₀₂ = 91814.857 oX ₀₂ = -40.689 oY ₀₂ = 88.798 oZ ₀₂ = -96.954

Table A4. Universal representation of the lung model, visualization of the alveolar model taking into account the phenomenon of collision
(continued from the previous page)

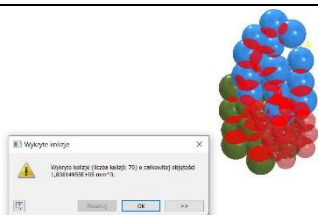
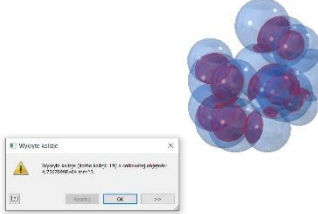
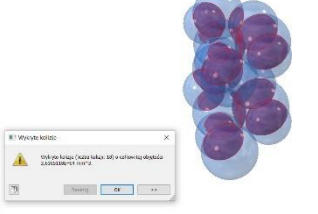
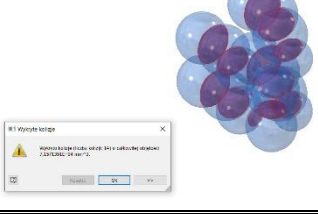
Right lung, ID = 35, visualization and parametrization of the collision phenomenon			
	ID = 35 WKEp ₃₅ = 051/048 VK ₃₅ = 1.83814955E+05 WKEV ₃₅ = 0.109611 WKEV ₃₅ = 10.96%	V ₃₅ = 1676965.655 P ₃₅ = 241930.985 oX ₃₅ = -9.812 oY ₃₅ = -76.728 oZ ₃₅ = -70.887	
	ID = 03 WKEp ₀₃ = 019/016 VK ₀₃ = 4.7567899E+04 WKEV ₀₃ = 0.069178 WKEV ₀₃ = 6.92%	V ₀₃ = 687611.770 P ₀₃ = 94180.514 oX ₀₃ = -5.645 oY ₀₃ = -69.274 oZ ₀₃ = -10.786	
	ID = 04 WKEp ₀₄ = 018/016 VK ₀₄ = 3.6105118E+04 WKEV ₀₄ = 0.131018 WKEV ₀₄ = 13.10%	V ₀₄ = 275572.550 P ₀₄ = 51103.633 oX ₀₄ = 34.763 oY ₀₄ = -83.086 oZ ₀₄ = -96.744	
	ID = 05 WKEp ₀₅ = 014/016 VK ₀₅ = 7.1571351E+04 WKEV ₀₅ = 0.100270 WKEV ₀₅ = 10.03%	V ₀₅ = 713781.335 P ₀₅ = 96646.839 oX ₀₅ = -31.036 oY ₀₅ = -81.453 oZ ₀₅ = -118.801	

Table A5. Summary table of quantitative and qualitative indicators in relation to the bubble model

Sets	Quantitative indicators	Qualitative indicators
ID = 15	WKGn _{15/15} = 100/080	WUOv _{15/15} = 1
	WKGv _{15/15} = 0.073873	WIWv _{15/15} = 0.073873
ID = 12	WKGn _{12/12} = 030/032	WUOv _{12/15} = 0.413985
	WKGv _{12/12} = 0.023283	WIWv _{12/15} = 0.002245
	WKGv _{12/15} = 0.005422	
ID = 01	WKGn _{01/01} = 011/016	WUOv _{01/12} = 0.436400
	WKGv _{01/01} = 0.019467	WUOv _{01/15} = 0.180663
	WKGv _{01/12} = 0.004779	WIWv _{01/12} = 0.002085
	WKGv _{01/15} = 0.001978	WIWv _{01/15} = 0.000357
ID = 02	WKGn _{02/02} = 016/016	WUOv _{02/12} = 0.563600
	WKGv _{02/02} = 0.024336	WUOv _{02/15} = 0.233322
	WKGv _{02/12} = 0.007715	WIWv _{02/12} = 0.004348
	WKGv _{02/15} = 0.003194	WIWv _{02/15} = 0.000745
ID = 35	WKGn _{35/35} = 051/048	WUOv _{35/15} = 0.586015
	WKGv _{35/35} = 0.109611	WIWv _{35/15} = 0.021174
	WKGv _{35/15} = 0.036132	
ID = 03	WKGn _{03/03} = 019/016	WUOv _{03/35} = 0.410033
	WKGv _{03/03} = 0.069178	WUOv _{03/15} = 0.240286
	WKGv _{03/35} = 0.01595	WIWv _{03/35} = 0.006540
	WKGv _{03/15} = 0.00935	WIWv _{03/15} = 0.002247
ID = 04	WKGn _{04/04} = 018/016	WUOv _{04/35} = 0.164328
	WKGv _{04/04} = 0.131018	WUOv _{04/15} = 0.096299
	WKGv _{04/35} = 0.012111	WIWv _{04/35} = 0.001990
	WKGv _{04/15} = 0.007097	WIWv _{04/15} = 0.000683
ID = 05	WKGn _{05/05} = 014/016	WUOv _{05/35} = 0.425639
	WKGv _{05/05} = 0.100270	WUOv _{05/15} = 0.249431
	WKGv _{05/35} = 0.024007	WIWv _{05/35} = 0.010218
	WKGv _{05/15} = 0.014068	WIWv _{05/15} = 0.003509

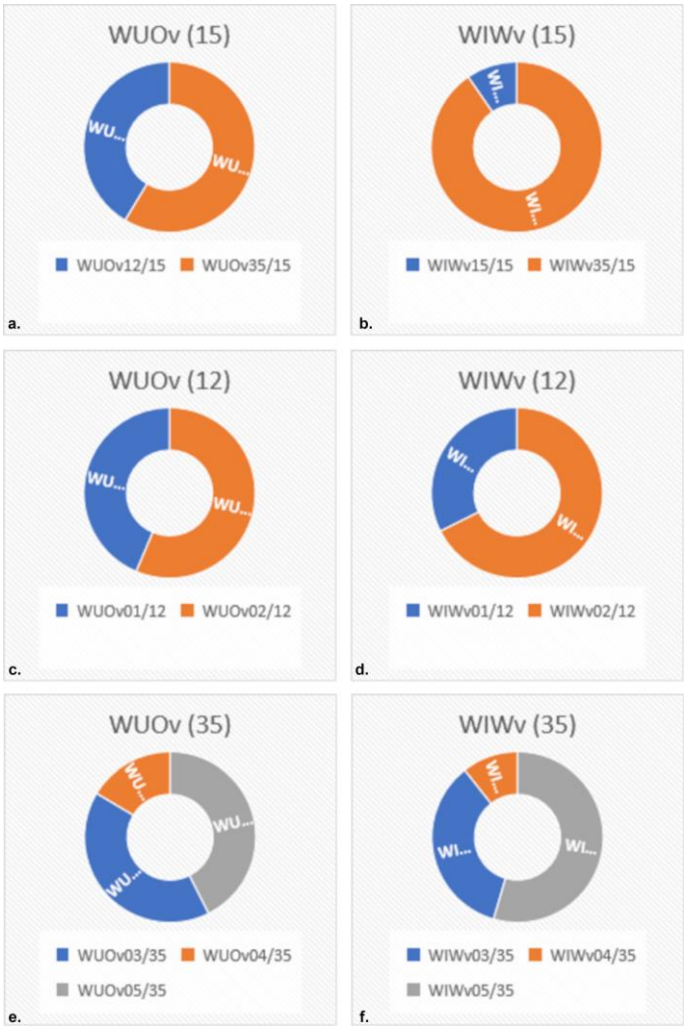


Figure A1. Donut charts representing: a,b - the share and quantitative impact of individual sets ID 12, 35 on the structural elements of the higher order ID: 15; c,d - share and quantitative impact of individual sets ID 01, 02 on structural elements of the higher order ID: 12; e,f - share and quantitative impact of individual sets ID 03, 04, 06 on structural elements of a higher order ID: 35