

CT-based lung geometry for high-resolution modelling of cardiopulmonary interaction

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Abstract

Background: Accurate representation of lung geometry is a prerequisite for reliable computational modelling of cardiopulmonary interaction, particularly for simulating regional ventilation–perfusion (V/Q) heterogeneity driven by gravity and vascular anatomy. Most existing models either simplify lung geometry or lack explicit spatial anchoring to physiological reference points.

Material and methods: In this study, diagnostic computed tomography (CT) images were used to reconstruct anatomically faithful lung geometry and subdivide both lungs into 80 spatially coherent regions. For each region, a set of geometric parameters was extracted, including volume and three-dimensional coordinates of the centre of gravity relative to the pulmonary trunk and to horizontal and vertical reference planes. Image preprocessing, vectorization, and geometric reconstruction were performed using widely available software tools, and the resulting feature matrix was implemented in a mathematical model of cardiopulmonary interaction within a virtual patient environment.

Results: The CT-based approach enabled precise localization of the pulmonary trunk and consistent definition of a global coordinate system, allowing simultaneous representation of vertical (gravitational) and radial (vascular resistance–related) perfusion gradients. Example simulations demonstrated physiologically plausible distributions of ventilation and perfusion across lung regions, including reduced ventilation in dependent areas and heterogeneous perfusion within iso-gravitational planes.

Conclusions: CT-based reconstruction of lung geometry from diagnostic images enables high-resolution regional partitioning and generation of a robust geometric matrix suitable for cardiopulmonary simulations. Although limited to a single anatomical dataset and static geometry, the presented model provides a solid foundation for future integration of regional mechanics, dynamic deformation, and validation with functional imaging, supporting advanced research, educational, and clinical applications.

Keywords: CT-based lung reconstruction; geometric parameters; modelling; cardiopulmonary interaction

Introduction

Cardiopulmonary interaction and regional V/Q heterogeneity

There are numerous direct interactions between the cardiovascular and respiratory systems, both of which are also influenced by external factors. Among these, gravity plays a predominant role,

as it affects all individuals and modulates both pulmonary perfusion and ventilation. Consequently, the ventilation–perfusion (V/Q) ratio—one of the key determinants of oxygen delivery to tissues and carbon dioxide removal—varies across lung regions due to gravitational effects. In particular, perfusion in the upper lung regions is relatively lower than ventilation, whereas the lower lung regions are better perfused than ventilated^{1,2}.

Pulmonary perfusion refers to the flow of blood through the pulmonary capillary network supplying the alveoli, the sites of gas exchange. Together with ventilation—the movement of air into and out of the lungs—it underpins effective oxygen uptake and carbon dioxide removal. Regional pulmonary perfusion is determined not only by gravitational hydrostatic pressure gradients related to vertical lung position, but also by vascular resistance associated with the distance from the pulmonary trunk^{3,4}.

In silico cardiopulmonary modelling: rationale and current limitations

Advanced in vivo analysis of the interaction between the cardiovascular and respiratory systems in both health and disease—such as chronic obstructive pulmonary disease or heart failure, which are among the leading causes of death^{5,6}—is extremely challenging. This difficulty arises from the complexity of the mutual interactions between the two systems as well as the influence of multiple external factors. Consequently, in silico analyses currently represent the most feasible approach, at least for preliminary investigations of such a complex physiological interplay. For this reason, a virtual cardiopulmonary patient has been developed^{7–17}.

To the best of the authors' knowledge, existing models typically describe the mechanics of the respiratory and/or cardiovascular systems with varying degrees of precision, but do not explicitly account for lung geometry, or they incorporate lung geometry while simulating mechanical properties only in a simplified manner^{18–23}. Moreover, the original virtual patient developed by the authors²⁴ represented the lungs as only five lobes, which proved insufficient for simulating regional differences in ventilation–perfusion (V/Q) relationships.

Aim of the study

For these reasons, the aim of this study is to subdivide the lungs into multiple anatomically and spatially coherent regions, each defined by explicit geometric coordinates relative to gravitational reference planes and the pulmonary trunk, thereby enabling

physiologically grounded simulation of regional ventilation and perfusion heterogeneity. This article describes a new variant of a lung partitioning method developed for modelling pulmonary circulation and used in simulations of cardiopulmonary interaction (CPI) within a virtual patient (VP), extending the approach previously presented by Pałko et al⁴.

Materials and Methods

CT data acquisition and image preprocessing

The overall workflow for CT-based lung geometry preprocessing and vectorization is summarized in **Figure 1**. The primary source of data was a computed tomography examination (**Figure 1A**). Example anonymized tomographic images (Supplementary **Figure S1**) were obtained from the Central Clinical Hospital of the Ministry of Internal Affairs and Administration. The image resolution was 5 mm, covering the full vertical extent of the lungs.

Tomographic images contain extensive anatomical detail, much of which is not required for computational modelling. Constructing a reliable lung model requires only an accurate outline of the lungs and a reference point that enables alignment of consecutive slices. Nevertheless, the use of tomographic imaging provides the high spatial accuracy necessary to reconstruct precise geometric models. The use of Autodesk Inventor allows you to import raster and vector images into a 2D sketch, especially in the *.dwg format, which is a standard in the description of vector objects.

The images were initially prepared using the GiMP raster image processing software and subsequently subjected to contour tracing with the Potrace algorithm, resulting in binary lung parenchyma segmentation (**Figure 1B**). The extracted contours were then vectorized to generate two-dimensional representations of the lung envelope (**Figure 1C**). The final format of files prepared for import in Autodesk Inventor is the universal vector format *.dxf.

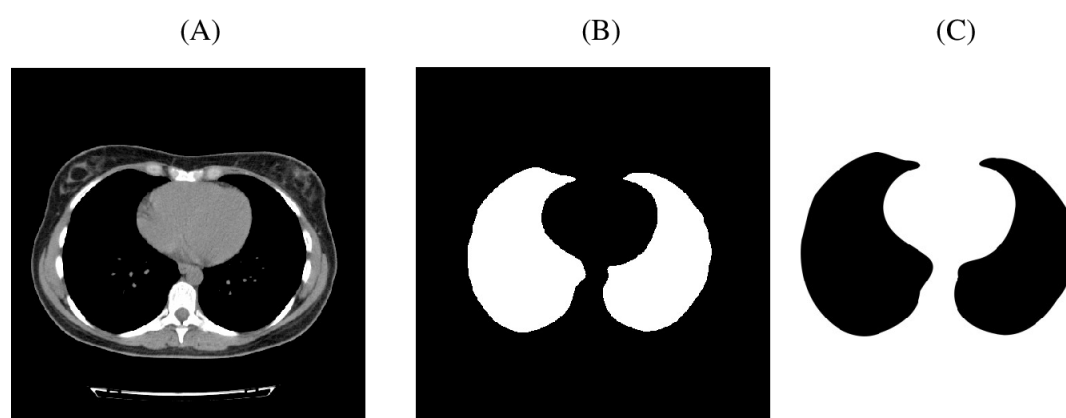


Figure 1. CT-based lung geometry preprocessing and vectorization workflow.

- (A) Representative axial CT slice of the thorax (slice thickness 5 mm) used as source data for lung geometry reconstruction.
- (B) Binary image obtained after preprocessing of the CT slice (lung parenchyma segmentation), used for positioning of cross-sections and extraction of lung contours.
- (C) Two-dimensional vector representation of the lung envelope obtained after contour extraction and vectorization, serving as the geometric boundary for subsequent 3D reconstruction.

Geometric model structure and coordinate system definition

The model structure consists of a stack of sketches with relative spacing consistent with the resolution of the acquired tomographic images (**Figure 2**). In this case, the inter-slice spacing was 5 mm. Each sketch contains a single imported vector boundary.

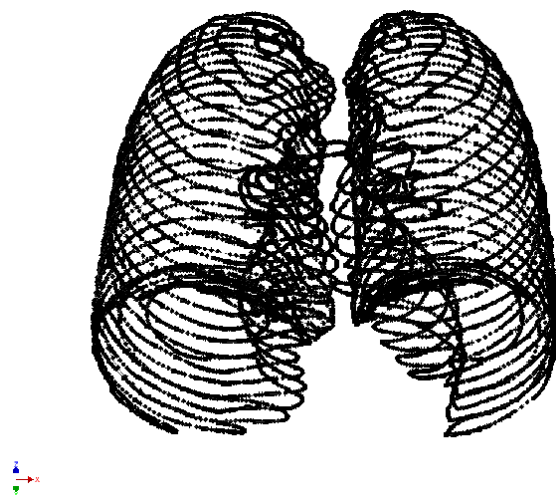


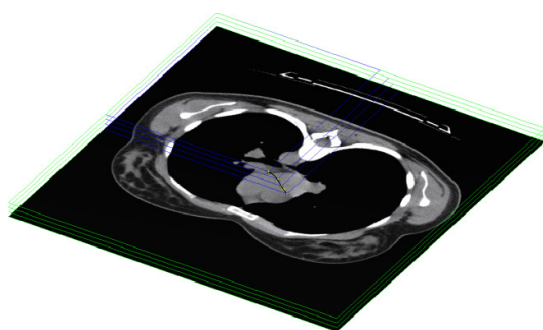
Figure 2. Stack of vectorized lung envelopes.

Stack of vectorized lung envelopes imported as parallel sketches and arranged according to the spacing of consecutive CT slices (5 mm). The stacked structure forms the geometric basis for subsequent 3D lung reconstruction and spatial parameter extraction.

The location of the pulmonary trunk centre was determined based on tomographic images. To achieve more accurate localization, its position was identified in multiple cross-sections as the centre of an inscribed circle. The designated centres were then connected using a spatial spline line (**Figure 3**). For model positioning, the origin (0,0,0) of the world coordinate system (WCS), which serves as the reference frame for calculating the model's physical properties, was aligned with this spline.

Lung reconstruction and lobar segmentation strategy

Lung reconstruction was performed using two alternative approaches, as illustrated in **Figure 4**. The first approach employed straight extrusion of consecutive cross-sections, resulting in a characteristic stepped surface between adjacent slices



(**Figure 4A**). This method allows reconstruction of cross-sections of arbitrary geometric complexity and preserves the exact contours extracted from tomographic images.

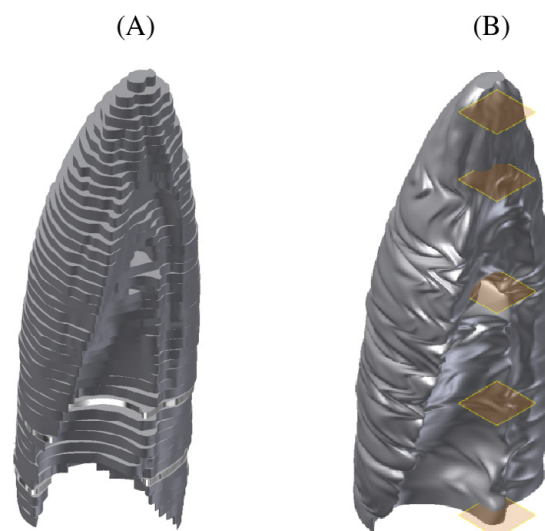


Figure 4. Comparison of slice-based and loft-based lung reconstruction methods.

(A) Three-dimensional lung geometry reconstructed by straight extrusion of consecutive vectorized cross-sections derived from computed tomography images. Each cross-section is extruded independently with inter-slice spacing corresponding to the CT acquisition resolution (5 mm), resulting in a characteristic stepped surface between adjacent sections while preserving the exact cross-sectional contours and spatial consistency.

(B) Three-dimensional lung geometry reconstructed using a loft-based technique that interpolates smooth surfaces between consecutive cross-sections. Although this approach produces visually smooth surface transitions, it requires substantial simplification of cross-sectional geometry and proved sensitive to segmentation complexity, leading to degradation of the reconstructed volume when the lungs were split into separate closed solids.

The second approach used a loft-based technique to generate smooth transitional surfaces between consecutive sections (**Figure 4B**). However, this method requires substantially simplified cross-sectional geometries to ensure surface continuity.

In practice, the loft-based reconstruction led to degradation of the model when the lung volume was split from a single solid into two separate closed loops during the reconstruction process, preventing generation of a stable and fully smooth

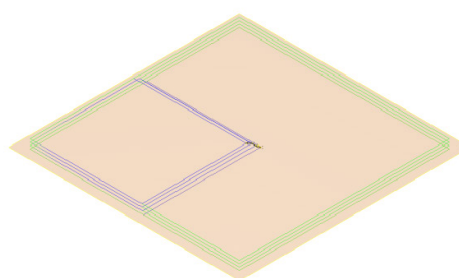


Figure 3. Localization of the pulmonary trunk.

Determination of the pulmonary trunk location based on tomographic images. In consecutive cross-sections, the centre of the pulmonary trunk was identified as the centre of an inscribed circle. The resulting centres were connected using a spatial spline, which served as the reference axis for positioning the model within the world coordinate system (WCS).

model. Consequently, the slice-based straight extrusion method was selected, as it ensured methodological consistency and robustness of the resulting lung geometry.

Data from an anatomical atlas were used to determine the surfaces for division into lobes (**Figure 5**). The method applied in the present work follows the approach developed in the earlier version of the model (Variant I) and remains consistent with the methodology described by Pałko et al.⁴ For each lung region, geometric parameters were extracted, including regional volume and three-dimensional coordinates of the centre of gravity defined relative to the pulmonary trunk and to horizontal and vertical reference planes. These parameters were implemented as input arrays in a mathematical model of cardiopulmonary interaction.

Each lung region is uniquely defined by a four-element identifier ID $[n, j, k, m]$, where n denotes the lung lobe, j the cranio-caudal position, k the radial position relative to the pulmonary trunk, and m the anteroposterior location. For every region, the three-dimensional coordinates of the centre of gravity (COG; x, y, z) and the corresponding regional volume were extracted. The complete numerical feature matrix is provided in Supplementary Table S1.

Results

Three-dimensional lung geometry and regional partitioning

CT-based reconstruction yielded anatomically coherent three-dimensional geometries of both lungs. Application of the hierarchical segmentation strategy (see the example of the right lung in **Figure 6**) resulted in subdivision of the lung volumes into 80 spatially consistent regions distributed across all lobes.

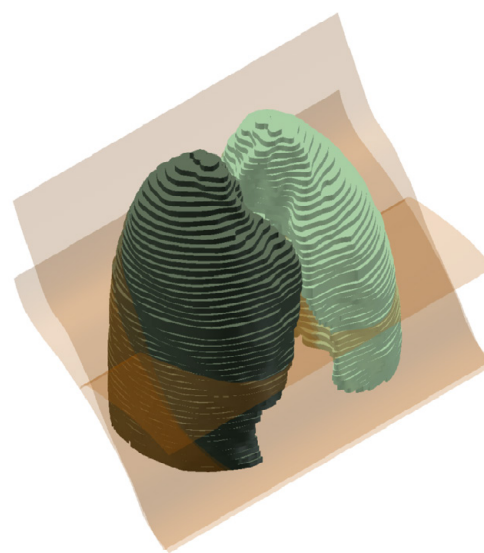


Figure 5. Method for division of the lungs into anatomical lobes. Visualization of the reconstructed left and right lung volumes with superimposed geometric division planes used to delineate anatomical lobes. The division surfaces were defined based on anatomical atlas data and applied consistently across the reconstructed lung geometry. This step provides the structural framework for subsequent regional segmentation and extraction of lobe-specific geometric parameters.

Each region constituted a contiguous anatomical volume with well-defined spatial boundaries. A complete visualization of the right-lung segmentation hierarchy is provided in Supplementary Figure S2.

For all regions, geometric parameters were successfully obtained, including regional volume and three-dimensional coordinates of the centre of gravity defined relative to the pulmonary trunk as well as to horizontal and vertical reference planes.

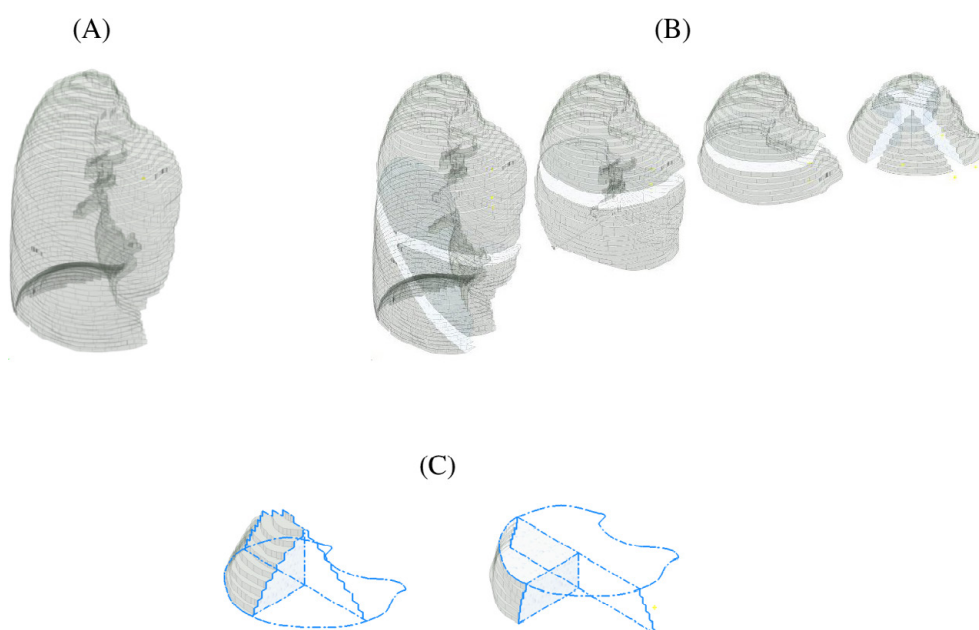


Figure 6. Segmentation strategy of the right lung.

- (A) Three-dimensional geometry of the reconstructed right lung
 (B) Stepwise subdivision of the lung volume into progressively smaller regions based on predefined anatomical and geometric planes.
 (C) Example of a single fourth-level segment illustrating the applied partitioning planes used to define spatial subregions.

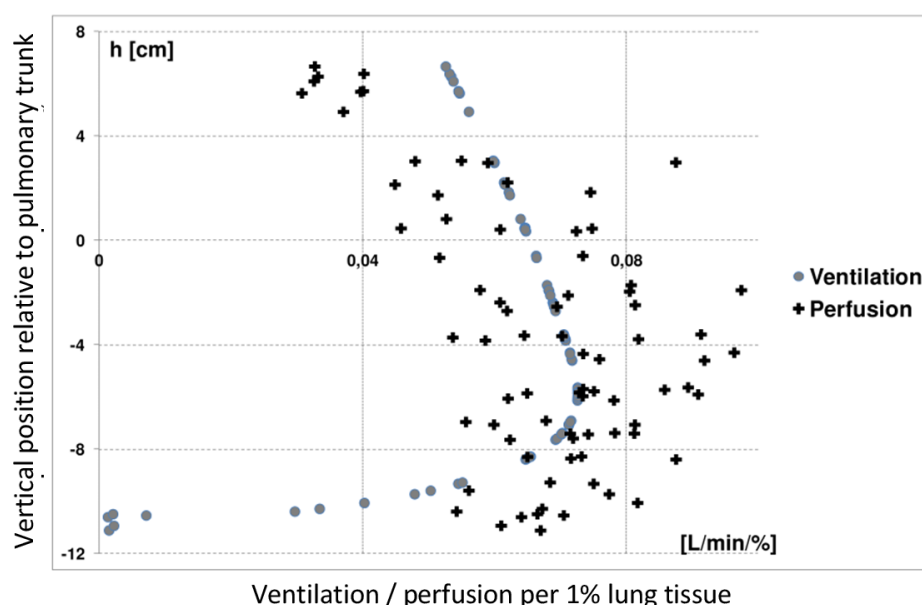


Figure 7. Influence of lung region location on resting ventilation and perfusion in a virtual patient.

Regional ventilation (circles) and perfusion (crosses) are shown as flow per 1% of lung tissue plotted against vertical position (h) of the regional centre of gravity relative to the pulmonary trunk.

Total ventilation and perfusion were 6.0 and 6.5 L/min, respectively. Most dependent regions exhibit low or absent ventilation during tidal breathing due to closing capacity exceeding functional residual capacity, whereas perfusion varies both with vertical position and radial distance from the pulmonary trunk, resulting in heterogeneity among regions at similar gravitational heights.

(Simulation performed using the VirRespir virtual patient environment³⁰).

Spatial reference system and extracted geometric parameters

Identification of the pulmonary trunk enabled definition of a global spatial reference system aligned with the central vascular axis of the lungs. This framework allowed simultaneous characterization of regional position along vertical (gravitational) and radial (pulmonary trunk-based) axes across the entire lung volume.

The resulting dataset provided a complete, spatially resolved geometric description of all lung regions, suitable for direct implementation in subsequent cardiopulmonary modelling analyses.

Feature matrix of lung regions

The extracted geometric parameters were assembled into a feature matrix comprising all 80 lung regions. The matrix includes region identifiers, volumes, and centre-of-gravity coordinates and serves as a compact representation of lung geometry suitable for direct use in cardiopulmonary simulations. Detailed numerical values are provided in Supplementary Table S1.

Example simulation of regional ventilation and perfusion distribution

An example simulation was performed using the generated geometric matrix to illustrate regional distributions of ventilation and perfusion within a virtual patient. The simulation demonstrated marked heterogeneity across lung regions.

Ventilation was reduced or absent in the most dependent regions during tidal breathing, while perfusion exhibited variability both along the vertical axis and among regions located

at similar vertical positions. Regions at comparable gravitational height showed differences in perfusion associated with their spatial relationship to the pulmonary trunk (Figure 7).

Discussion

The present study demonstrates that lung geometry extracted directly from diagnostic CT images enables a detailed and spatially coherent division of the lungs into 80 functional regions. This significantly advances the resolution of cardiopulmonary simulations compared with classical models, in which the lungs are represented by only a few compartments per lung^{25,26}. Previous approaches to cardiopulmonary modelling typically focused either on the mechanical interactions of the respiratory and cardiovascular systems – often with simplified geometric assumptions—or on anatomically realistic reconstructions without a comprehensive description of mechanical or perfusion behavior^{18–23}. The present work bridges this gap by developing a geometrically explicit and computationally usable matrix of lung regions for use within a virtual patient model.

A major advantage of the CT-based model is its high anatomical fidelity, including precise positioning of the pulmonary trunk and accurate determination of regional centres of gravity. These parameters are essential, as pulmonary perfusion depends not only on the vertical location of a given region (hydrostatic pressure gradient) but also on the radial distance from the pulmonary trunk (vascular resistance gradient)^{3,4}. While vertical gradients of V/Q are well described in classical physiology and experimental imaging studies, radial perfusion

heterogeneity within iso-gravitational planes is far less commonly represented in computational simulations despite its physiological significance^{25,27}.

This behaviour is illustrated in **Figure 7** and arises from fundamental physical differences between blood and air. Because blood viscosity is orders of magnitude higher than that of air, pulmonary perfusion is influenced not only by hydrostatic pressure gradients related to vertical lung position, but also by vascular resistance associated with radial distance from the pulmonary trunk^{28,29}. Consequently, lung regions located at similar gravitational heights may exhibit markedly different perfusion values.

An important source of difference between the CT-based lung geometry presented in this study and atlas-based lung models lies in the consistency and origin of the input data used for model construction. The CT-based model derives directly from patient-specific imaging and therefore provides substantially higher geometric accuracy and internal coherence, particularly with respect to spatial relationships between lung regions, the pulmonary trunk, and gravitational reference planes. This precision, however, comes at the cost of generalizability, as the resulting geometry represents an individual anatomical case rather than a population-average structure. In contrast, atlas-based lung models offer broader applicability by averaging anatomical features across multiple subjects and prior observations, but this averaging inherently reduces geometric fidelity and may obscure subject-specific spatial gradients relevant to perfusion and ventilation.

Crucially, the CT-derived approach enables precise localization of the pulmonary trunk and an unambiguous definition of the global coordinate system (0,0,0 WCS), which directly determines regional distances used in the geometric matrix. As pulmonary perfusion is sensitive to both gravitational height and radial distance from the central vascular axis, accurate spatial anchoring of these reference points is essential for reliable representation of perfusion gradients. Atlas-based models, while valuable for generalized simulations, lack this level of positional specificity and therefore provide less consistent regional input for high-resolution cardiopulmonary simulations.

The workflow presented here – combining image preprocessing (GIMP), vectorization (Potrace), and geometric modelling (Autodesk Inventor) – demonstrates that CT-based reconstruction can be performed with accessible tools while maintaining the geometric precision necessary to support simulations of perfusion and ventilation distribution. The decision to use a “stepped” extrusion method rather than a smoothed loft approach ensures methodological consistency and avoids errors caused by segmentation discontinuities between left and right lungs. Although the resulting surfaces are not perfectly smooth, the method preserves the robustness of the regional subdivision, which is crucial for downstream computational modelling.

An example simulation of ventilation and perfusion distribution illustrates that the spatial segmentation is physiologically meaningful: dependent lung regions demonstrate low ventilation due to their closing capacities exceeding functional

residual capacity in this specific case, while perfusion varies with both height and proximity to the pulmonary trunk. This pattern is consistent with established physiological knowledge and prior imaging studies.

Despite these strengths, the current study focuses primarily on geometric reconstruction. Integration of regional mechanics, validation with functional imaging, and extension to dynamic simulations represent natural next steps. Nevertheless, the presented geometry forms a necessary foundation for future development of more advanced virtual cardiopulmonary models, with potential applications in research, education, and clinical decision support.

Several limitations of the present work should be acknowledged. First, the model is based on a single CT dataset, which restricts its generalizability. Inter-individual variability in lung anatomy, thoracic shape, and vascular tree configuration may influence regional geometry and thereby affect perfusion and ventilation distributions. Second, the current model focuses solely on static geometry. Regional mechanical properties—including non-linear compliance, spatial distribution of airway resistance, and gravitational deformation—were not incorporated. As such, while the geometric matrix allows for realistic perfusion gradients, mechanical behaviour of parenchyma remains simplified. Third, the “stepped” reconstruction method provides robustness but lacks the smooth surface continuity achievable through loft-based methods. Although the choice was deliberate to maintain segmentation consistency, minor geometric discontinuities between slices remain. Finally, the model has not yet undergone functional validation using imaging modalities such as EIT, SPECT/CT, PET, or dual-energy CT. Such validation is essential to confirm the physiological accuracy of predicted regional ventilation and perfusion patterns.

Future development of this model may proceed in several directions:

1. **Population-based geometric modelling**
Extending the method to multiple CT datasets would enable creation of averaged, probabilistic, or parametric lung geometries capturing anatomical variability across populations.
2. **Integration of regional mechanics**
Coupling geometric segmentation with spatial distributions of compliance, resistance, recruitment thresholds, and airway tree topology would significantly enhance physiological realism.
3. **Dynamic deformation modelling**
Implementing deformable meshes or biomechanical finite-element methods could allow simulation of respiratory motion and time-varying changes in perfusion and ventilation.
4. **Functional imaging validation**
Comparisons with EIT, SPECT/CT, or dual-energy CT perfusion maps would provide essential validation and could help refine the geometric partitioning or regional parameter weighting.
5. **Clinical and educational applications**
The model may be incorporated into virtual clinical training platforms, pre-operative planning tools, or simulation

environments evaluating mechanical ventilation strategies, body positioning, or disease-induced heterogeneity.

6. Automation of segmentation

Implementation of AI-based segmentation and automated centre-of-gravity extraction would streamline replication of the method and allow rapid development of patient-specific models.

Conclusions

This study presents a lung model derived directly from diagnostic CT images and subdivided into 80 anatomically consistent regions, forming a geometric matrix suitable for simulations of cardiopulmonary interaction. CT-based segmentation enables accurate spatial anchoring of lung regions relative to both gravitational reference planes and the pulmonary trunk, which are key determinants of regional ventilation and perfusion.

Although based on a single dataset and limited to static geometry, the proposed approach represents a methodological step toward anatomically grounded computational lung modelling. By linking high-resolution imaging with mathematical simulation within a virtual patient environment, it provides

a foundation for future extensions incorporating regional mechanics, dynamic deformation, and validation with functional imaging.

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Conflicts of interest

Authors of the manuscript “*CT-based lung geometry for high-resolution modelling of cardiopulmonary interaction*” state that there are no Conflicts of Interest to declare.

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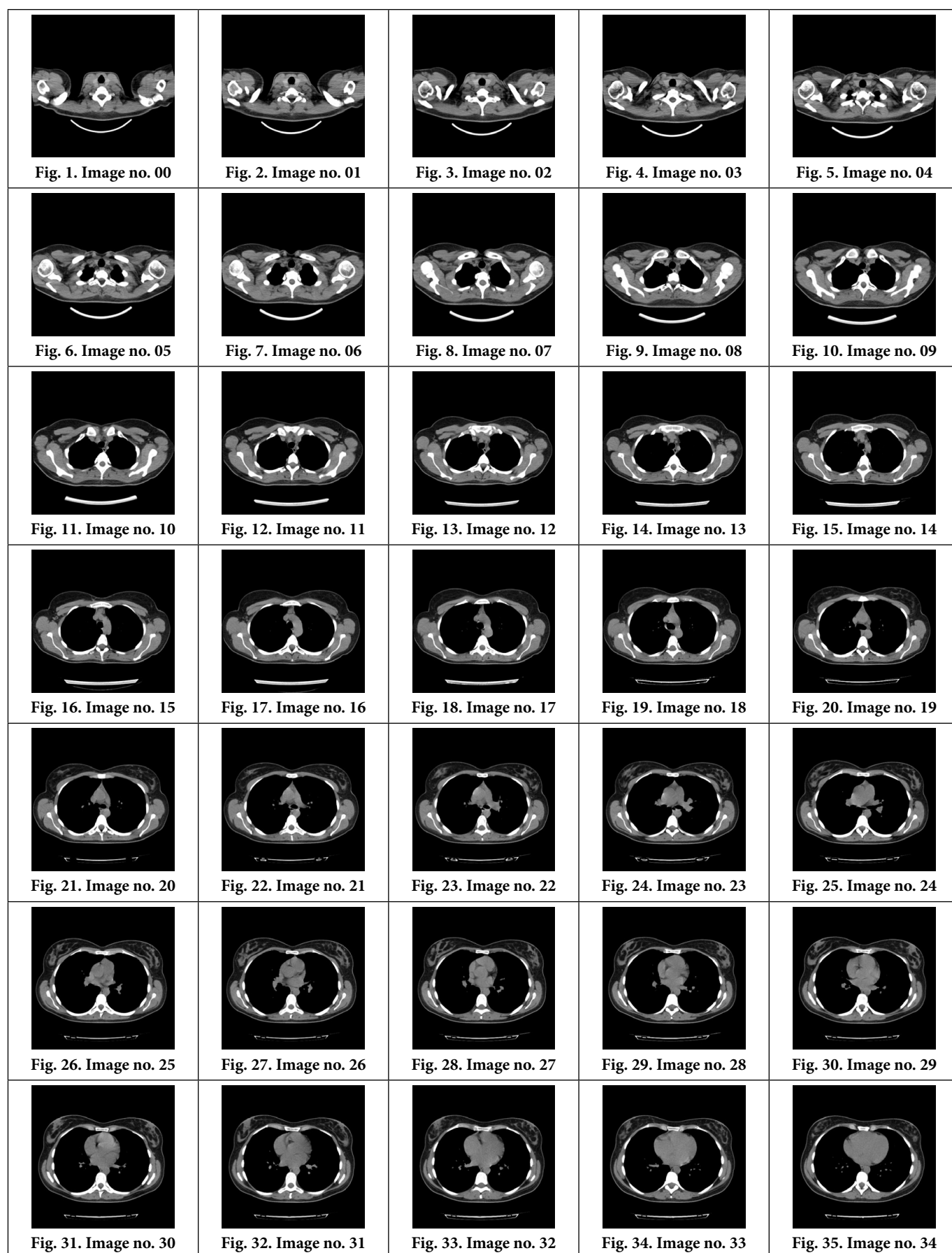
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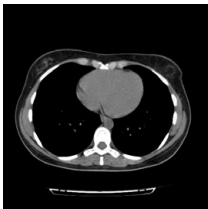
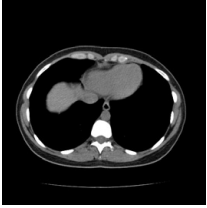
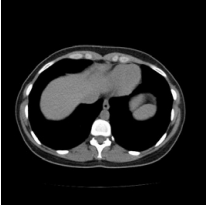
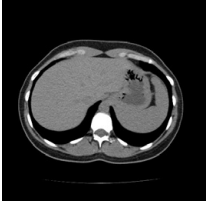
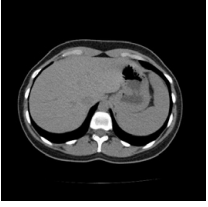
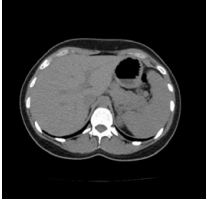
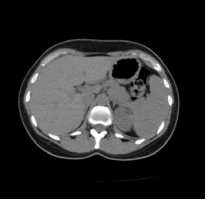
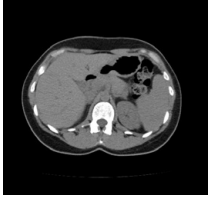
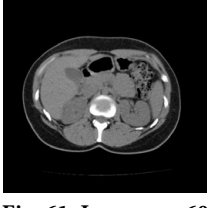
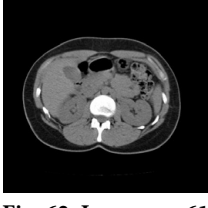
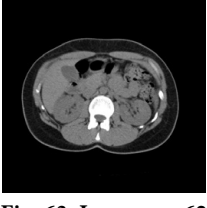
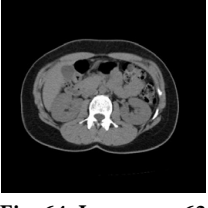
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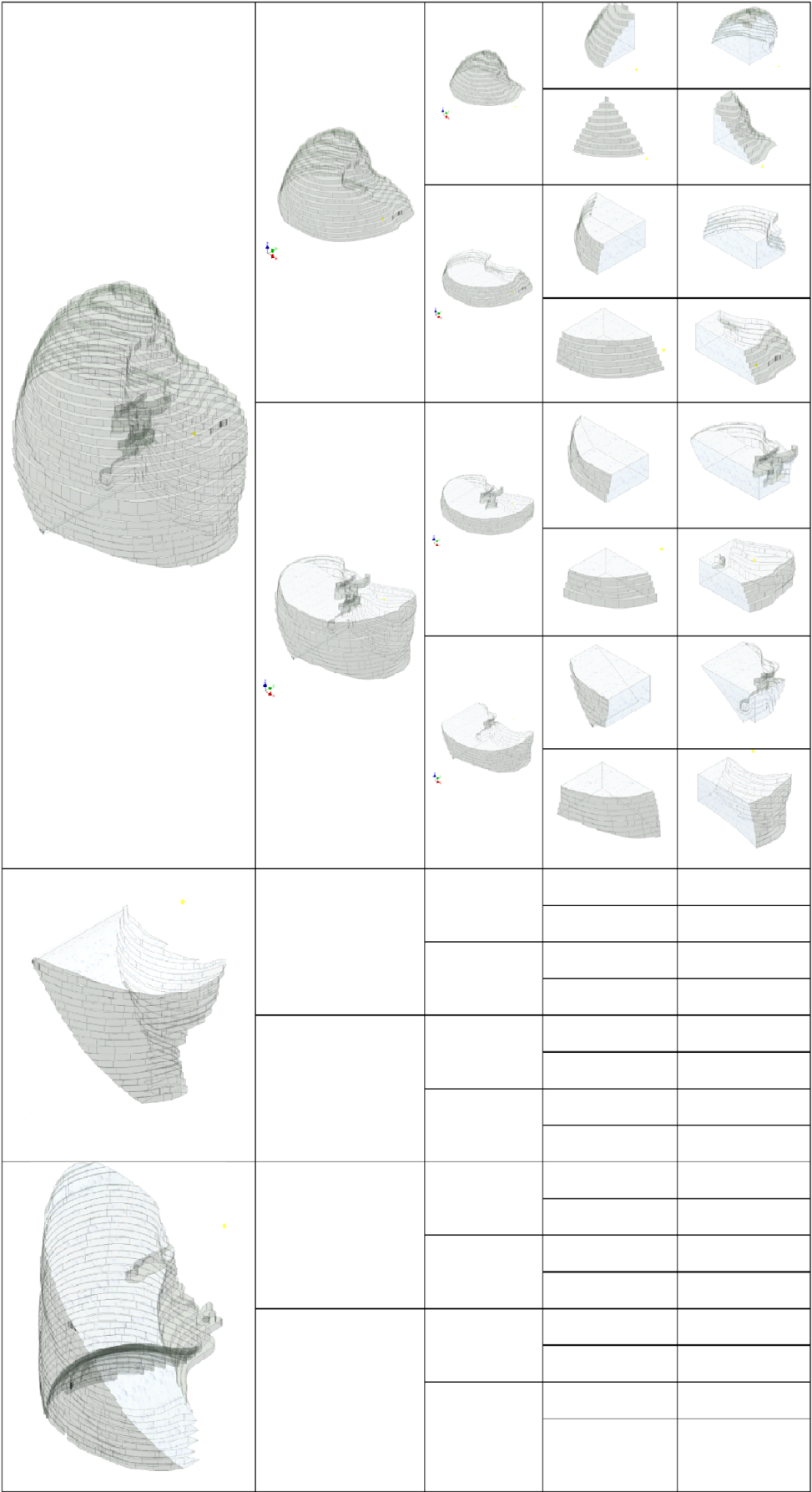
Appendix A. Supplementary data



Supplementary Figure S1. Example anonymized tomographic images (cross-sections) taken at 5 mm resolution.

 Fig. 36. Image no. 35	 Fig. 37. Image no. 36	 Fig. 38. Image no. 37	 Fig. 39. Image no. 38	 Fig. 40. Image no. 39
 Fig. 41. Image no. 40	 Fig. 42. Image no. 41	 Fig. 43. Image no. 42	 Fig. 44. Image no. 43	 Fig. 45. Image no. 44
 Fig. 46. Image no. 45	 Fig. 47. Image no. 46	 Fig. 48. Image no. 47	 Fig. 49. Image no. 48	 Fig. 50. Image no. 49
 Fig. 51. Image no. 50	 Fig. 52. Image no. 51	 Fig. 53. Image no. 52	 Fig. 54. Image no. 53	 Fig. 55. Image no. 54
 Fig. 56. Image no. 55	 Fig. 57. Image no. 56	 Fig. 58. Image no. 57	 Fig. 59. Image no. 58	 Fig. 60. Image no. 59
 Fig. 61. Image no. 60	 Fig. 62. Image no. 61	 Fig. 63. Image no. 62	 Fig. 64. Image no. 63	

Supplementary Figure S1 (cont.).



Supplementary Figure S2. Full hierarchical subdivision of the right lung into geometrically defined region.
The figure presents the full hierarchical subdivision of the reconstructed right lung volume into progressively smaller geometrically defined regions.

Supplementary Table S1. Feature matrix of CT-derived lung regions.

The table presents geometric parameters of 80 anatomically coherent lung regions obtained from diagnostic computed tomography (CT). For each region, the matrix includes a hierarchical region identifier, three-dimensional coordinates of the centre of gravity relative to the pulmonary trunk and reference planes, and regional volume.

n = 1...5 {1: left-top, 2: left-bottom, 3: right-top, 4: right-middle, 5: right-bottom}

j = 0...1 {0: up, 1: down} {00, 01, 10, 11}

k = 0...1 {0: centripetal, 1: lateral}

m = 0...1 {0: front, 1: rear}

ID [n, j, k, m]						COG [x, y, z]			
Lp	n	j	k	m	X	Y	Z	V	
Left lung – upper lobe {CT method}									
1	1	1	00	0	0	-11.636	52.556	57.285	48584.802
	2	1	00	0	1	-49.531	49.046	62.850	113368.919
	3	1	00	1	0	-11.897	86.488	49.376	29862.651
	4	1	00	1	1	-51.900	88.160	56.456	79095.812
	5	1	01	0	0	10.082	50.690	18.537	85549.073
	6	1	01	0	1	-45.806	47.061	22.282	58261.488
	7	1	01	1	0	-0.368	94.122	17.412	99220.965
	8	1	01	1	1	-45.227	98.079	21.491	71717.615
	9	1	10	0	0	37.192	59.293	-19.497	71893.692
	10	1	10	0	1	-4.671	68.785	-17.074	66051.341
	11	1	10	1	0	31.349	99.889	-23.689	31126.202
	12	1	10	1	1	-5.731	109.220	-18.968	95587.109
	13	1	11	0	0	43.280	66.163	-74.208	87264.414
	14	1	11	0	1	7.825	74.434	-61.229	20415.022
	15	1	11	1	0	35.252	103.231	-76.273	70577.913
	16	1	11	1	1	7.693	110.362	-60.502	44781.364
Left lung – lower lobe {CT method}									
2	1	2	00	0	0	-50.986	52.654	-5.842	40103.128
	2	2	00	0	1	-81.359	59.784	8.282	86198.367
	3	2	00	1	0	-50.129	105.891	-6.573	37803.667
	4	2	00	1	1	-79.380	96.730	4.725	58793.211
	5	2	01	0	0	-43.028	56.814	-37.767	67858.434
	6	2	01	0	1	-81.164	60.436	-36.413	54411.401
	7	2	01	1	0	-39.638	108.636	-38.271	89521.682
	8	2	01	1	1	-80.549	100.825	-37.163	54563.514
	9	2	10	0	0	-27.363	59.152	-70.560	77211.476
	10	2	10	0	1	-71.784	62.986	-69.005	71089.597
	11	2	10	1	0	-23.353	114.302	-70.508	86109.268
	12	2	10	1	1	-71.640	108.487	-69.493	66115.277
	13	2	11	0	0	-29.862	47.328	-97.167	52730.222
	14	2	11	0	1	-71.946	62.551	-105.940	78974.728
	15	2	11	1	0	-15.666	125.825	-95.818	54179.747
	16	2	11	1	1	-73.591	110.590	-103.811	73504.250
Right lung – upper lobe {CT method}									
3	1	3	00	0	0	-16.713	-19.600	63.897	54648.091
	2	3	00	0	1	-55.020	-19.179	66.722	73095.951
	3	3	00	1	0	-18.151	-52.114	57.040	32747.366
	4	3	00	1	1	-56.502	-53.106	61.035	55955.281
	5	3	01	0	0	1.011	-16.353	29.961	71927.296
	6	3	01	0	1	-63.988	-19.874	30.608	53826.712
	7	3	01	1	0	-8.844	-59.439	29.775	71160.127
	8	3	01	1	1	-60.480	-59.337	30.412	69795.324

Supplementary Table S1 cont.

ID [n, j, k, m]					COG [x, y, z]				
Lp	n	j	k	m	X	Y	Z	V	
Right lung – upper lobe {CT method}									
	9	3	10	0	0	15.297	-24.173	4.949	70693.753
	10	3	10	0	1	-58.515	-25.749	4.646	80937.293
	11	3	10	1	0	5.405	-67.632	3.580	48667.354
	12	3	10	1	1	-50.276	-69.761	4.237	76559.419
	13	3	11	0	0	22.637	-31.874	-24.167	68453.106
	14	3	11	0	1	-43.230	-28.096	-18.991	45534.891
	15	3	11	1	0	10.997	-68.934	-24.685	76098.587
	16	3	11	1	1	-39.263	-73.524	-20.986	60468.620
Right lung – middle lobe {CT method}									
	1	4	00	0	0	25.758	-50.078	-45.944	9506.565
	2	4	00	0	1	-12.566	-49.892	-42.902	15015.387
	3	4	00	1	0	19.792	-77.638	-45.463	6457.644
	4	4	00	1	1	-11.595	-83.567	-43.398	16851.751
	5	4	01	0	0	35.953	-50.272	-57.178	15860.395
	6	4	01	0	1	-5.271	-56.180	-56.291	6248.652
	7	4	01	1	0	21.510	-77.783	-57.692	10629.647
	8	4	01	1	1	-4.397	-83.932	-56.818	11738.854
	9	4	10	0	0	41.338	-55.777	-73.754	9770.797
	10	4	10	0	1	12.819	-61.585	-73.889	8429.119
	11	4	10	1	0	32.468	-76.598	-75.803	3621.182
	12	4	10	1	1	12.516	-83.899	-73.958	14877.998
	13	4	11	0	0	39.130	-60.097	-105.317	13436.614
	14	4	11	0	1	21.550	-62.180	-93.157	2521.075
	15	4	11	1	0	33.806	-77.014	-102.760	6897.324
	16	4	11	1	1	21.202	-83.034	-92.693	4525.771
Right lung – lower lobe {CT method}									
	1	5	00	0	0	-53.416	-24.254	-35.975	21612.521
	2	5	00	0	1	-84.186	-25.179	-25.331	66850.030
	3	5	00	1	0	-51.628	-73.090	-36.563	25561.633
	4	5	00	1	1	-81.045	-64.057	-26.954	41990.993
	5	5	01	0	0	-46.154	-26.047	-58.942	43659.605
	6	5	01	0	1	-81.381	-25.222	-58.205	42543.954
	7	5	01	1	0	-37.946	-74.963	-59.623	71101.388
	8	5	01	1	1	-79.538	-65.100	-58.513	31060.948
	9	5	10	0	0	-28.105	-27.577	-83.837	47269.203
	10	5	10	0	1	-72.207	-28.103	-82.741	58568.725
	11	5	10	1	0	-19.530	-78.962	-83.491	67110.228
	12	5	10	1	1	-69.187	-71.793	-82.881	40986.863
	13	5	11	0	0	-28.272	-20.330	-100.513	22260.912
	14	5	11	0	1	-76.126	-27.006	-111.039	64803.824
	15	5	11	1	0	-7.240	-86.494	-104.809	47536.276
	16	5	11	1	1	-74.210	-73.435	-109.279	42111.994