Patient-specific Morphological and Blood Flow Analysis of Pulmonary Artery in the Case of Severe Deformations of the Lung due to Pneumothorax*

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Abstract

Pneumothorax is characterized by lung collapse. Its effect on hemodynamics, especially on pulmonary arterial blood flow, remains unclear. This patient-specific study investigated the effects of lung deformation on pulmonary blood flow during acute phase and after recovery. Arterial geometry was extracted up to the fifth generation from computed tomography images in three patients and reconstructed. Different geometrical parameters (artery bores, area ratios, and between-branch angles) were computed. The shapes of the pulmonary trunk and its branches were affected strongly by pneumothorax. To clarify the effect of geometrical perturbations on blood flow, the Navier–Stokes equations for a steady laminar flow of Newtonian incompressible fluid were solved in a reconstructed domain. The change in flow structure between acute phase and recovery was associated with variations in flow rate ratio between the right and left lungs. This study shows, possibly for the first time, that from a patient-specific numerical test, pneumothorax has a considerable impact on pulmonary arterial morphology and hemodynamics.

Key words: Pneumothorax, Computational Fluid Dynamics, Morphological Analysis, Pulmonary Arteries, Patient-specific Model

1. Introduction

Pneumothorax is the presence of air between the parietal and visceral pleura\(^1\). It can occur spontaneously and causes lung collapse. It was first recognized in 1803 by Jean Marc Gaspard Itard and described in 1819 by Rene Laennec\(^2\). As pneumothorax alters the geometry of the lung parenchyma, which includes deformable blood vessels, it disturbs blood flow, particularly in the pulmonary arteries. Among the different types of pneumothorax, secondary spontaneous pneumothorax is a complication of preexisting lung disease and affects mainly people over 60 years old\(^3\). This potentially life-threatening event results from not only reduced respiratory reserves\(^4\) but also impaired pulmonary circulation. Air leaks can persist for more than 2 weeks after secondary spontaneous pneumothorax\(^5\).

It has been shown in sheep that for right-sided pneumothorax, the blood flow and
conductance in the right lung are reduced markedly, whereas no evident impact is observed in the left lung\(^5\). In humans, Breconsky described the impact of pneumothorax on blood circulation in 1930\(^5\). Since then, many studies have been undertaken, mostly in animal. Leigh-Smith and Harris have provided a good overview of these studies\(^7\).

A major outcome of pneumothorax is alveolar hypoxia and hypoxemia\(^8\). Hypoxia is known to generate hypoxic pulmonary vasoconstriction (HPV) in the pulmonary arterial microcirculation\(^9\). HPV is aimed at redirecting blood flow to the alveoli with higher oxygen tension\(^10\). However, pneumothorax-induced collapse targets all blood vessels of whatever type or size.

Previous studies have shown that it takes 3–4 days to revert the pulmonary blood flow to normal distribution after pneumothorax\(^11\). Yamazaki et al. found a decrease of 66–75% in pulmonary blood flow in lungs affected by pneumothorax just after re-expansion\(^11\). Carvalho et al. observed an 80% decrease in pulmonary arterial blood flow in the affected lung\(^5\). However, these clinical publications have not yielded information on pulmonary artery morphology and provide even less data on blood flow behavior. Pneumothorax is an acute disorder that requires rapid investigation of structural and functional aspects. As three-dimensional imaging is a common practice nowadays, numerical tests can be performed to exhibit blood flow behavior in the pulmonary artery during and after the acute phase.

Various models of arterial blood flow\(^12\) and morphometry of pulmonary arteries\(^13\) have been developed. Some surgical and interventional procedures have significant failure rates, which indicate a need to study the fluid dynamics before and after intervention for optimal planning\(^14\). Since the late 1990s, the number of patient-specific computational hemodynamic studies has increased considerably\(^15\). Some researchers have even developed an open source hemodynamics computational fluid dynamics (CFD) project\(^16\). To the best of our knowledge, there has been no numerical study of the impact of pneumothorax on pulmonary arterial circulation. Furthermore, persistent consequences on the pulmonary arterial morphology and blood flow have not been studied carefully.

The present study compared differences in the geometry of the pulmonary artery based on computed tomography (CT) images acquired during the acute disease phase and after recovery, provided by Tohoku University Hospital. Possibly for the first time, blood flow simulation in a patient-specific domain reconstructed from image processing demonstrated tissue morphology changes and their effects on blood flow.

2. Method

2.1 Materials

We studied two CT image sets acquired during the acute phase of pneumothorax and after a short or long delay, from each of three patients aged over 60 y. 1) A male patient, referred to as m1, who suffered simultaneously from pneumothorax in the right lung and liver injuries. Bloody ascites was suspected. The slice interval was 2 mm. The delay between the acquisitions of the two image sets was 2 days. 2) A male patient, referred to as m2, who suffered simultaneously from pneumothorax in the left lung and mediastinal leakage. The CT image slice interval was 1 mm. Image set 2 was acquired 77 days after set 1. 3) A female patient, referred to as f1, who suffered simultaneously from pneumothorax in the right lung and a suspected thrombus in the pulmonary arteries [A3 and A7 branches for the right pulmonary artery (RPA), and A8 branch for the left pulmonary artery (LPA)]. She was also affected by mesothelioma. The slice intervals were 1 and 2 mm in image sets 1 and 2, respectively. Image set 2 was acquired 88 days after set 1.
For all patients, an Aquilion 64-detector row helical CT scanner (Toshiba, Tokyo, Japan) was used. Scans were obtained with the following parameters: 0.5 second per rotation, 0.5 mm collimation, and 55 mm/s table increment (pitch=55, Beam pitch=0.859). Patients were requested to breath in and hold their breath for approximately 8 seconds during the scanning. To improve the delineation of pulmonary arteries, 90 ml of non-ionic contrast material was injected at a rate of 1 ml s⁻¹ via the antecubital vein in each patient. When the attenuation value at the level of the main pulmonary artery reached a preset threshold (CT attenuation value of 100 HU plus initial CT value of the pulmonary artery), helical scanning automatically started. Transverse sections were reconstructed with a 1-mm section thickness at 1-mm intervals. Theoretical spatial resolution was 0.5mm in XY-plane and 1mm in Z-axis.

Written informed consent for contrast-enhanced CT scanning was obtained from the patient. Because this CT examination was routine clinical examination using our standard CT protocol, our IRB waved its approval for this retrospective study.

Figures 1 and 2 illustrate the procedures used to extract one cross-section from another. They show the center of the respective cross-sections, R_{max} and R_{min} are the maximum and minimum distance, respectively, between C_i and the nodes of cross-section k.

2.2 Basic Equations for Analysis of Arterial Geometry

Many techniques have been used to build models of anatomical organs from noisy data sets. These data sets usually yield huge numbers of points and possible artifacts, even sometimes holes and contour intersections. Each technique has its advantages and drawbacks. Here, we used commercial software (MIMICS®, Materialise NV, Leuven, Belgium) for image segmentation and geometric 3D reconstruction. Image resolution and noise often do not allow one to distinguish the pulmonary arteries and veins as well as the arterial branches from the adjacent arteries, especially as vessel size decreases. Therefore, 3D reconstruction was limited to the main trunk (generation 0, G0) and down to G4, except for two G5 branches, one in each lung (Fig. 1).

A Fortran 77 (F77) code devoted to geometric feature analysis was developed, based on a centerline generation algorithm devised by Pivello et al. As shown in Fig. 2, by starting from the inlet and moving downstream, this algorithm permitted extraction of a series of cross-sections.

The procedure steps were as follows: 1) creation of a surface unstructured mesh using MIMICS; 2) tagging of inlet nodes to obtain the first cross-section; 3) for each section k, assessment of vessel section center coordinates and main geometric features; 4) selection of already unvisited surface elements that bound the section k; and 5) extraction of nodes of the selected elements that were not in the cross-section k, to draw the cross-section k+1. If the number of nodes in the cross-section was null, the corresponding
loop was stopped. Otherwise, we went back to step 3).

From MIMICS, for each case, we extracted three files: one for the main trunk geometry, one for the RPA, and one for the LPA. The former algorithm was applied for each file.

To analyze each case, the following coefficients were calculated for a cross-section \( k \) that contained \( n \) nodes. Let \( R_{ik} \) be the distance between the center \( C_k \) of the cross-section and the nodes \( i \). The coefficients \( R_{\text{max}}(k) \), \( R_{\text{min}}(k) \), and \( R_{\text{mean}}(k) \) were, respectively, the maximum, minimum, and mean of \( R_{ik} \).

An aspect ratio-like coefficient, \( \lambda(k) \), was defined as follows:

\[
\lambda(k) = \frac{R_{\text{max}}(k)}{R_{\text{min}}(k)}
\]

The standard deviation (SD) \( \sigma_R(k) \) of \( R_{ik} \) in percentage was defined as follows:

\[
\sigma_R(k) = \frac{10^4}{n} \times \sqrt{\frac{\sum_{i} (R_{\text{mean}}(k) - R_{ik})^2}{n}}
\]

A reference node \( P_{S2} \), at the top peak point of the second-generation branches’ split, was chosen as shown in Fig. 3. We calculated the distance from the center of each cross-section to this node for recovered and sick stages. \( C_{M,1} \) and \( C_{B,1} \) were the center of the first selected cross-section of the main trunk and one of the branches, respectively. To insure comparison of the values of the former coefficients in cross-sections at a similar position for sick and recovered conditions, we used the following distances:

\[
D_{M}(k) = -\left[ C_{M,k} - P_{S2} \right]
\]

\[
D_{B}(k) = \sum_{j=2}^{n} \left[ C_{B,j} - C_{B,j-1} \right]
\]

where \( D_{M} \) was the distance for the main trunk, and \( D_{B} \) was the distance for one branch. Note that to express the fact that \( D_{M} \) was calculated from the inlet to \( P_{S2} \), as observed in Fig. 2, we used a negative sign. We non-dimensionalized these distances, because each patient had different physiology. As we aim at comparing the acute and after recovery phases, and for the sake of simplicity, the bigger value of \( R_{\text{max}}(1) \) for the sick and recovered conditions was taken as a characteristic length, \( R_{\text{cyl}} \), and \( D_{M} \) and \( D_{B} \) were non-
dimensionalized as $\hat{D}_M(k) = D_M / R_{cyl}$ and $\hat{D}_B(k) = D_B / R_{cyl}$.

To compare the angle between the main trunk centerline and the left and right branch centerlines, we determined the three angles $\theta_1$, $\theta_2$, and $\theta_3$, as shown in Fig. 4. To calculate these angles, we first introduced three vectors, $\mathbf{u}_M$, $\mathbf{u}_L$, and $\mathbf{u}_R$, which were the direction vectors of the main trunk, left branch, and right branch centerlines, respectively, close to their junction. We calculated these vectors for two different lengths in order to ensure the result. For case 1, the vector length was about 5 mm, and the distance between the end points of these vectors and $P_{S2}$ was about 10 mm. For case 2, the vector length was about 10 mm, and the distance between the end points of these vectors and $P_{S2}$ was about 20 mm. $\mathbf{u}_L'$ and $\mathbf{u}_R'$ were the respective projections of $\mathbf{u}_L$ and $\mathbf{u}_R$ in the plane normal to $\mathbf{u}_M$. Then, we defined the following angles: $\theta_1$ was the angle $(\mathbf{u}_M, \mathbf{u}_L)$, $\theta_2$ was the angle $(\mathbf{u}_M, \mathbf{u}_R)$, and $\theta_3$ was the angle $(\mathbf{u}_L', \mathbf{u}_R')$.

### 2.3 Basic Equations for Analysis of Arterial Geometry

In order to discuss the effect of geometry change on the flow field, we performed CFD analysis under a simplified condition. We assumed that the blood was an incompressible Newtonian fluid, given that the non-Newtonian effect of blood is assumed to be small for a relatively large Reynolds number flow. Moreover, it could still be supposed that the erythrocyte aggregation time constant remained higher than the local flow time scale. We also assumed that the flow was laminar and steady, though the actual blood flow is time-dependent and may show slight turbulence transition in its decelerated period. These complexities may change the CFD results, however, data on the frequency content of the input signal in patients were lacking. Especially, we do not know how the pulsation of blood flow is modified by pneumothorax. Thus, in this study, we employed simplified CFD conditions and focus only on the geometry effect. The governing equations were given by the continuity equation (5) and the Navier–Stokes equation (6).

\[ \nabla \cdot \mathbf{v} = 0 \quad (5) \]
\[ \rho (\mathbf{v} \cdot \nabla) \mathbf{v} = -\nabla p + \mu \Delta \mathbf{v} \quad (6) \]

where $\mathbf{v}$ was the velocity of the flow; $p$, the pressure; $\rho$, the density; and $\mu$, the viscosity. We assumed that the patients had a normal hematocrit, and the density was consequently equal to 1060 kg m$^{-3}$. The blood viscosity is more than three times the water viscosity under high shear conditions and thus was fixed arbitrarily at $4 \times 10^{-3}$ Pa.s.

### 2.4 Computational Grid and Numerical Methods for CFD

Owing to the coarse slice intervals in the m1 and f1 cases, the computational analysis was performed only for m2. To reduce the computational load, we removed the small arteries and generated a mesh of up to 19 outlets for sick and recovered conditions: eight in the left lung and 11 in the right. The remaining branches were the same in both conditions, and the distance between an outlet and the last bifurcation for the same branch was almost the same, within an error of less than 5%, for sick and recovered conditions. We used a tetrahedral unstructured mesh, generated with GRIDGEN (Pointwise Inc., Fort Worth, TX, USA), FLUENT (Ansys Inc., Pittsburgh, PA, USA), a flow solver based on the finite volume method, was used to solve Eqs (5) and (6). The mesh size was similar between the sick and recovered conditions. Wang et al. have previously demonstrated the accuracy of unstructured tetrahedral mesh created with FLUENT for modeling blood flow in the case of total cavopulmonary connections. Short length: 30.8 mm in acute phase, 30.0 mm after recovery; diameter: 39 mm) straight tubes have been added in modeling the ends to ensure exit cross-sections, to avoid pressure gradients within entry and exit sections, to take into account upstream and downstream effects of 3D flow, and to eliminate numerical artifacts caused by the application of boundary conditions too close to the investigated domain.

At the inlet, the flow should have uniform velocity distribution, except near the wall.
that gave rise to a boundary layer with its high near-wall vorticity. The flow rate was \( Q = 1.91 \times 10^{-4} \text{ m}^3 \text{ s}^{-1} \) for the recovered condition. For the sake of simplicity, we used a flat velocity profile at the inlet. This condition is not far, anyway, from experimental results \((23, 24)\). As pneumothorax may affect the cardiac output \((5)\), three different conditions, \(0.5 \times Q\), \(Q\), and \(1.5 \times Q\), were set for the sick conditions. The cylinder inlet diameter was \( R_{cyl} = 19.5 \text{ mm} \) for both conditions, which means an average velocity at the inlet of 0.16 m/s and a Reynolds numbers at the original inlet of 1300 for the sick condition with \(Q\) (650 for \(0.5 \times Q\) and 2050 for \(1.5 \times Q\)) and 1270 for the recovered conditions, as we wanted to avoid instabilities due to a high Reynolds number. At the outlet, zero pressure was given. We applied a no-slip boundary condition to the wall. The convergence was considered reached when the residual for the continuity equation became lower than \(10^{-5}\).

We compared a mesh with about 100,000 and 350,000 tetrahedral cells. By comparing the average inlet pressure, we found less than 1% difference in the results. The chosen mesh contained 99741 cells or the sick stage and 97517 cells for the recovered stage.

![Figure 5. Front (a, c) and side (b, d) views of sick-stage (a, b) and recovered (c, d) pulmonary arteries. The arrow shows a bump in the main trunk in the sick stage.](image)

![Figure 6. Inlet for the (a) recovered and (b) sick conditions.](image)

### Table 1. Comparison of \(\theta_1\), \(\theta_2\), and \(\theta_3\) between the sick and recovered conditions.

<table>
<thead>
<tr>
<th></th>
<th>(\theta_1)</th>
<th>(\theta_2)</th>
<th>(\theta_3)</th>
</tr>
</thead>
<tbody>
<tr>
<td>m2</td>
<td>Recovered</td>
<td>46.8°</td>
<td>54.7°</td>
</tr>
<tr>
<td></td>
<td>Sick</td>
<td>53.5°</td>
<td>59.8°</td>
</tr>
<tr>
<td>f1</td>
<td>Recovered</td>
<td>64.3°</td>
<td>57.3°</td>
</tr>
<tr>
<td></td>
<td>Sick</td>
<td>81.7°</td>
<td>54.9°</td>
</tr>
<tr>
<td>m1</td>
<td>Recovered</td>
<td>78.2°</td>
<td>46.2°</td>
</tr>
<tr>
<td></td>
<td>Sick</td>
<td>86.8°</td>
<td>30.5°</td>
</tr>
</tbody>
</table>

### 3. Results

#### 3.1 General Observations

The CT images depicted obvious differences in the shape of the lungs between image sets 1 and 2; therefore, we expected to also find differences in the shape of the pulmonary arteries. Figure 5 shows the images for case m2, under sick conditions and after recovery. A simple comparison of the arterial geometry between the two conditions in the front (Fig. 5a and c) and side (Fig. 5b and d) views led us to hypothesize that the pulmonary arterial geometry was affected strongly by the occurrence of the disorder. We also observed, as
shown by the arrow in Fig. 5b, a bump in the main artery in the sick condition that did not appear after recovery. We noted a more obvious difference in the shape of the inlet (Fig. 6). After recovery, it was shaped somewhere between a square and a circle, whereas in the sick condition, it looked like an elongated ellipsoid.

3.2 Split Angles of Second Generation Branches

Table 1 compares the angles $\theta_1$, $\theta_2$, and $\theta_3$ between the sick and recovered conditions. There was an increase in $\theta_3$ after recovery in all three patients, while $\theta_1$ and $\theta_2$ did not show any significant change between the two conditions.

3.3 Changes in the Cross-sectional Shape

Figure 7 shows the values of $\sigma_R$ for cross-sections at different positions in the main trunk during the sick stage (blue circle) and after recovery (red triangle). The value of $\sigma_R$ was slightly larger during the sick stage in patient m2 (Fig. 7c). In patients f1 (Fig. 7a) and m1 (Fig. 7b), a similar, although less pronounced, situation was observed, while the cross-section center $C_k$ became closer to the node $P_{S2}$ at the junction between the RPA and LPA. These results indicated that severe deformation of the lung led to increased irregularity of the main trunk’s cross-sectional shape. For the LPA and RPA, the results varied among the patients; thus, we could not draw a general conclusion regarding the deformation of cross-sectional shape in these arteries.

The validity of the cross-sectional cut was checked with the coefficients $\sigma_\alpha$ and $\beta$. $\sigma_\alpha$ was always less than $3^\circ$, and $\alpha$ was less than $1^\circ$, which means that the cross-section’s normal vector was close enough to the centerline. For all patients, and for the main trunk, RPA and LPA, the coefficient $\lambda$ had similar results to $\sigma_R$. It was confirmed by computing two other coefficients similar to $\lambda$, with $R_{\text{mean}}$ instead of $R_{\text{min}}$ for one and instead of $R_{\text{max}}$ for the other, which also showed similar results (data not shown).

3.4 Ratio of Cross-sectional Area between Left and Right Branches

Finally, we compared the ratio of cross-sectional areas between the RPA and LPA close to $P_{S2}$, by selecting cross-sections with centers at the same two distances from $P_{S2}$: about 10 and 20 mm. As both lengths gave similar results we don’t show the results for the smallest length here. Due to arterial wall motion, we didn’t consider the absolute cross-sectional value of any importance, so that here we present only the cross-sectional area ratio. Table 2 shows the results for a distance of 20 mm between the cross-section center and $P_{S2}$. The ratio of the RPA to the LPA was the same before and after recovery, one-third in patient m2 and two-thirds in m1. For patient f1, the ratio differed by about 10% between the sick and recovered stages (70:30 vs. 60:40, for sick vs. recovered). Although the reason for this difference was not clear, there might have been a natural difference in the proportion of the cross-sectional area between the right and left second generation branches in this patient, or a thrombus might have been present in a pulmonary artery.

Table 3 shows the ratio of the cross-sectional areas of the left and right outlets of the meshed geometry for the sick and recovered conditions in patient m2, as used for the CFD study. For the fourth-generation arteries, which are situated inside the lungs, the ratio showed a significant difference between the sick and recovered conditions. The outlet cutting procedure under MIMICS can generate small differences in outlet position or shape, but this should not create an error bigger than 1% and cannot explain the difference in the outlet area ratio shown in Table 3.
Table 2. Ratio of cross-sectional areas of the right (S_R) and left (S_L) second-generation branches.

<table>
<thead>
<tr>
<th></th>
<th>S_R (%)</th>
<th>S_L (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>m2</td>
<td>Recovered</td>
<td>65%</td>
</tr>
<tr>
<td>m2</td>
<td>Sick</td>
<td>67.5%</td>
</tr>
<tr>
<td>m1</td>
<td>Recovered</td>
<td>63.5%</td>
</tr>
<tr>
<td>m1</td>
<td>Sick</td>
<td>68%</td>
</tr>
<tr>
<td>f1</td>
<td>Recovered</td>
<td>60%</td>
</tr>
<tr>
<td>f1</td>
<td>Sick</td>
<td>68.5%</td>
</tr>
</tbody>
</table>

The ratio of cross-sectional areas of the right (S_R) and left (S_L) second-generation branches of the pulmonary artery in the sick and recovered conditions. The areas were calculated at about 20 mm from the split peak position, P_S2. The grey cells indicate branches running to the affected lung.

3.5 CFD Analysis

There was a pressure loss of about 444 Pa between the inlet and outlet in the computational domain for the sick stage, with a flow rate of Q, Re=1300, at the inlet (about 970 Pa for 1.5×Q, Re=2050, and 111 Pa for 0.5×Q, Re=650), compared with 363 Pa for the recovered stage. This might have been caused by a change in the outlet cross-sectional area. In both stages, high pressure was observed where the main flow from the inlet collided with the bifurcation around P_S2.

Figure 8 shows the velocity streamlines after recovery, Re=1270 (Fig. 8a), and in the sick condition, Re=1300 (Fig. 8b). We observed fewer streamlines in the LPA than in the RPA after recovery (Fig. 8a), whereas there was no difference in streamlines between the RPA and LPA during the sick condition (Fig. 8b), suggesting a more balanced flow in the sick stage compared with after recovery. We observed vortexes in area E_1 and E_2 (Fig. 8a) in the recovered stage and in area E_1', E_2' and E_3 (Fig. 8b), close to the Y-junction between the RPA and LPA, in the LPA and in the bump, in the sick stage. There was an increase in maximum velocity from 0.73 m s⁻¹ in the recovered stage to 0.81 m s⁻¹ in the sick stage, for a flow rate of Q at the inlet (maximum velocity of 1.18 m s⁻¹ for 1.5×Q and 0.43 m s⁻¹ for 0.5×Q). Figure 9 shows the velocity vectors in the main branches after recovery (Fig. 9a) and in the sick stage (Fig. 9b) in order to allow a better visualization of local velocity.
Figure 8. Velocity streamlines. (a) Recovered stage. (b) Acute phase with $Q$. RPA: Right Pulmonary Arteries. LPA: Left Pulmonary Arteries. $E_1$, $E_1'$ and $E_3$ show vortexes close to the Y-junction, $E_2$ and $E_2'$ show vortexes in the LPA in the recovered and sick stages. Units for velocity are expressed in m/s.

Figure 9. Velocity vector. (a) Recovered stage. (b) Acute phase with $Q$. RPA: Right Pulmonary Arteries. LPA: Left Pulmonary Arteries. Units for velocity are expressed in m/s.
Figure 10 shows the wall shear stress in the main branches after recovery (Fig. 10a) and in the sick stage (Fig. 10b). The wall shear stress in the RPA was not significant different between the sick and recovered stages, but there was a clear decrease in wall shear stress in the LPA after recovery compared with the sick stage. The maximum wall shear stress decreased from 15.9 Pa in the sick stage to 12.9 Pa in the recovered stage, with a flow rate of $Q$ at the inlet (33.0 Pa for $1.5Q$ decreased to 4.9 Pa for $0.5Q$).

Finally, Table 4 shows the ratio of the flow rates between left and right artery between left and right artery outlets in the sick and recovered conditions. In the sick stage, the flow rates were almost balanced, with about 52% in the right outlet and 48% in the left outlet, with a flow rate of $Q$ at the inlet (51.5:48.5 for $0.5Q$ and 51.8:41.2 for $1.5Q$). The flow rate became unbalanced after recovery, with a ratio of 60% in the right outlet and 40% in the left outlet. These results match with the cross-sectional area ratio between the right and left outlets shown in the previous section and with the streamlines in Fig. 8.

Table 4. Flow rate balance between the right and left outlets.

<table>
<thead>
<tr>
<th>Volume Flow Rate Ratio</th>
<th>L.O.</th>
<th>R.O.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Recovered</td>
<td>40%</td>
<td>60%</td>
</tr>
<tr>
<td>Sick, $0.5Q$</td>
<td>48.5%</td>
<td>51.5%</td>
</tr>
<tr>
<td>Sick, $Q$</td>
<td>48.2%</td>
<td>51.8%</td>
</tr>
<tr>
<td>Sick, $1.5Q$</td>
<td>48%</td>
<td>52%</td>
</tr>
</tbody>
</table>

The flow rate balance was calculated between the left outlets (L.O.) and the right outlets (R.O.) for the sick and recovered conditions. The grey cells indicate branches running to the affected lung.

4. Discussion

We showed changes in the shape of the pulmonary arteries between the sick and recovered stages of pneumothorax. Especially in the main trunk, where $\sigma_R$ was higher on average (+1.12%, +2.36%, and +5.11% for patients f1, m1, and m2, respectively); $\lambda$ also showed an average increase (+0.086, +0.035, and +0.46 for patients f1, m1, and m2, respectively). Even though we did not observe such general tendencies in the LPA and
RPA, these findings indicate that the pulmonary arteries are affected outside the lung in patients with severe deformation such as that due to pneumothorax. To look for persistent effects of the disease, information about the morphology in these patients before they contracted pneumothorax is needed. However, it is very difficult to obtain CT images of healthy persons just before they contract pneumothorax.

Considering that this was a patient-specific study, differences in morphology would be expected. Even though drawing general results by comparing patients with different physiological conditions is not a trivial matter, we were able to show a bend in the Y-junction between the main trunk and the RPA and LPA outside the lung. This study shows that the shape of the pulmonary arteries is modified not only inside the affected lung but also outside the lung. These results must be confirmed by more patient-specific studies.

In the present study, we used a pressure equivalent to that at the outlet boundary in the CFD study. This was not far from the physiological condition, because the pressure loss between the first bifurcation and the outlet of the CFD mesh was much smaller than that for the microcirculation. The main difference between the present boundary condition and the actual blood flow in patient m2 is that the CFD model did not have any resistance at the outlet. In the actual artery-vein network, there is a large resistance in the capillaries that generates a large pressure loss. The Reynolds number of these capillaries is usually less than unity, i.e., Stokes flow, and the pressure loss is proportional to the flow rate. In particular, in the case of HPV, the resistance increases significantly and regulates the flow rate ratio between the RPA and LPA. This phenomenon was observed by Carvalho et al. (5). These resistance effects were not taken into account in the present study. As the effect of HPV has already been studied by many groups (25, 26, 27, 4, 8), we focused on the effect of pneumothorax in large arteries.

Among its effects on pulmonary arterial blood flow, HPV can increase arterial pressure (9). The effects of HPV on pulmonary circulation from the cellular point of view have already been demonstrated (6). Bartsch et al. have demonstrated a reduction in cross-sectional area of the pulmonary capillaries (25). Blood flow heterogeneity also has been found in some cases of HPV (5). Recently, Schwenke et al. have found changes in macrovessel pulmonary distribution by holding experiments on rats (8). However, it is unclear whether HPV changes the distribution of human pulmonary blood flow (27), and this needs to be examined in pulmonary arteries in future studies.

It is known from former clinical research that, for a healthy person, the RPA is longer and larger than the LPA (28). We also observed a difference in the cross-sectional area ratio between the two arteries, although the imbalance was reduced in the sick stage. Under the present outlet condition of CFD analysis, the cross-sectional area ratio has clear consequences for the blood flow rate ratio between the two lungs. By studying three different flow rate conditions for the flow field study, we observed a similar flow rate ratio. However, the cross-sectional area balance between the RPA and LPA at the fourth-generation branches may not play a major role in the flow rate ratio, because the large resistance, which determines the flow rate ratio, is usually dominated by the capillaries. We should note, however, that the present CFD results indicate a significant change in blood flow between the sick stage and recovery, even without considering the resistance change in the microcirculation. Thus, we believe that the pulmonary blood flow can be altered not only by changes in microcirculation but also by morphological changes in the large arteries caused by pneumothorax. We can also note that the volume flow rate of blood is considered, from clinical studies, greater in the right lung than in the left (29).

We also observed a bump, which caused an additional vortex, in the main trunk of patient m2 in the sick condition. Although we did not observe this feature in the other patients, this should be considered as a possible effect of severe lung deformation. This
type of local geometric change, as well as large-scale deformation of large arteries, may occur in patients with pneumothorax.

Even though this study may not contribute directly to the medical treatment of pneumothorax, it may, at least, provides information that may helps to characterize this disease, which is a necessary step in the way to conceive remedies to a disease.

5. Conclusion

By extracting the pulmonary arterial geometry from CT images, we demonstrated a change in the shape of the pulmonary arterial tree. In particular, the shape of the main trunk became more irregular, the cross-section became more elliptical, and the first pulmonary arterial Y-junction outside of the lung was bent. Computations under the chosen limited boundary condition (outlet: zero pressure, inlet: constant flow rate) showed an increase in the maximum pressure loss. We also observed a change in the blood volume flow rate ratio between the RPA and LPA, generated by the geometric changes and chosen boundary conditions in our system. These results demonstrate that pneumothorax has a considerable effect on pulmonary arterial morphology and pulmonary blood flow. This study shows, possibly for the first time in a patient-specific numerical study, how pneumothorax can affect pulmonary arterial hemodynamics.

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