Modeling, segmentation, and caliber estimation of bronchi in high-resolution computerized tomography

Françoise Prêteux*, Catalin I. Fetita*, and Philippe Grenier+

* Département Signal et Image, Institut National des Télécommunications,
9, rue Charles Fourier, 91011 Evry Cedex, France
+ Service de Radiologie, Hôpital de la Salpétrière, Paris, France

ABSTRACT

In this paper, we address bronchi segmentation in high resolution computerized tomography (HRCT) in order to estimate the bronchial caliber. The method developed is based on mathematical morphology theory, and relies on morphological filtering, marking techniques derived from the concept of connection cost, and conditional watershed. In order to evaluate the robustness of the segmentation and the accuracy of the caliber estimates, a realistic bronchi modeling based on physiological characteristics has been developed. According to the size of the bronchi, the accuracy is up to 90%. Results are presented and discussed.

Keywords: mathematical morphology, connection cost-based segmentation, bronchi modeling, bronchi caliber, HRCT images.

INTRODUCTION

Asthma is a widely spread respiratory disease, mainly in the industrialized countries. Important modifications in bronchi caliber occur in patients suffering from asthma. Such caliber variations appear during an attack or by voluntary inhalation of bronchial reactive substances (during a diagnose test or a therapy - see Figure 1).

Respiratory functional explorations are the traditional methods which reveal bronchi constriction and dilation. They cannot determine the location of the bronchi and their degree of morphological modification. In fact, the local morphological variations are completely overlooked at this medical examination which provides only a global evaluation with no location criteria.

At present, the high-resolution computerized tomography (HRCT) makes it possible to provide a much better visualization of bronchial structures and pulmonary arteries. Bronchial walls are accurately displayed, whatever the bronchus orientation with respect to the slice level. The smallest identified bronchial lumen corresponds to bronchus having a diameter of 1 mm (sixth order bronchus).

Recent studies have shown that the HRCT makes it possible to estimate the topography and the intensity of the bronchial reactivity for different stimuli in animal experimentations. Similarly, for human lungs, the normal or dilated aspect of the bronchial lumen is easily visualized for the majority of cases.

However, the diagnosis of the normal bronchial caliber is sometimes difficult. In fact, diagnosis is often limited to comparisons between the bronchus surface and the surface of its homologous artery. But the variations of these calibers depend on many parameters, ventilatory cycle, cardiac cycle, arterial pression, etc., and sometimes have opposite signs. Perturbations of gas exchanges, induced by the ventilation defect, can act upon the artery caliber to which the bronchus caliber can no longer be compared. This problem shows the necessity for a method which enables absolute measurements of the bronchial and arterial caliber during a ventilatory cycle.

- Further author information -
F. Prêteux: Email: preteux@int-evry.fr; WWW: http://www-sim.int-evry.fr/People/Preteux.html
C. I. Fetita: Email: fetita@int-evry.fr
This research aims at developing an analysis and quantification system, *in vivo* and non-invasive, which makes it possible to study the bronchi morphology and their homologous artery, their spontaneous or induced variations, monitored on HRCT images. Our contribution consists in setting up an automatic bronchi segmentation procedure and a bronchus caliber estimation method.

Section 1 presents the specific segmentation method developed from the mathematical morphology theory. The segmentation method combines morphological filtering, connection cost-based marking and conditional watershed. In Section 2, the validation of the segmentation method is addressed in order to evaluate the accuracy of the bronchus caliber estimates. Several approaches have been applied: a statistical validation on a representative bronchi sampling, an absolute validation using phantom-images and a bronchus modeling-based validation. Results are presented and discussed in Section 3.

1 BRONCHI SEGMENTATION METHOD

1.1 Physiological characteristics of bronchi

In the pulmonary parenchyma, bronchi have a tree structure whose trunk lies in the superior part of the lung. Consequently, it is possible to see big bronchi simultaneously with smaller ones in the lung periphery. Each bronchus may have an associated satellite artery whose diameter approximately equals the bronchus diameter. However, small bronchi may sometimes degenerate, and in this case, the satellite artery is no longer present.

The study was carried out on a set of 15 HRCT images of sheep lungs. This set includes a collection of 80 bronchus sections defined on 512 x 512 pixel matrices corresponding to 1 mm thick slices, and reconstructed by using two different fields of view: 131 mm and 262 mm. In such images, the shape of a bronchus section is more or less elliptic according to the degree of the orthogonality of the slice level to the bronchus axis. The size and the grey scale distribution of the bronchus axis depend on the area of the lung under study (Figure 1).

![CT images of sheep lungs](image)

**Figure 1:** CT images of sheep lungs:
1a: Normal lung.
1b: After intravenous injection of histamine: bronchi are dilated.

1.2 Segmentation method

The developed segmentation method relies on the characteristic properties of the bronchi in terms of image analysis, on their environment and on image *a priori* knowledge. Thus, we can note the following:

- bronchi occupy only a small part of the image;
- the image noise is not very significant. It is essentially due to the artifacts created by the cardiac movement;
• the bronchial lumen has a very low grey level (between 0 and 25);
• the bronchial wall is very dense, corresponding to high grey levels (between 150 and 200);
• the pulmonary texture presents medium grey levels (between 80 and 120);
• the contrast between the bronchial wall and the bronchial lumen appears as the highest contrast in the image;
• the pertinent bronchi are quasi-circular. The bronchi which are not orthogonal to the slice level or those which are branching-off are not significant for physicians;
• the diameter of the pertinent bronchi varies from 3 to 30 pixels;
• some bronchial lumens have homogeneity defects essentially due to the rebounds of the high-frequency filters used in the reconstruction step for enhancing the contours.

This set of remarks led us to adopt a segmentation strategy based on mathematical morphology, for which the first step corresponds to a non-linear filtering using an isotropic sequential alternated filter of size $1^{13}$.

The second step consists in a morphological marking$^{14,15}$ of the bronchial lumen. Different marking functions can be used, as the extraction of the regional minima under constraints using the algorithm of $r$-$h$ extrema$^{16}$, or the algorithm of $h$-domes$^{17}$. In both cases, it is necessary to specify the constraint parameters $(h = \text{height}, r = \text{slope})$ which is extremely difficult in practice because of the great inter- and intra-image variability. This leads us to apply a non-parametrical marking method based on the concept of connection cost$^{18}$.

We recall that the connection cost is defined as follows:

$$
\forall (x, y) \in \mathbb{R}^2 \times \mathbb{R}^2, x \neq y, \quad C_f(x, y) = \inf \{ \lambda \in \mathbb{R} / \overline{\delta}_{f,\lambda}(x, y) < +\infty \},
$$

where $f$ is a grey level function and $\overline{\delta}_{f,\lambda}$ denotes the geodesic distance relative to the threshold of $f$ at level $\lambda$.

Schematically, the connection cost corresponds to the potential minimum barrier that must be crossed in order to pass from $x$ to $y$ (Figure 2).

![Figure 2: Connexion cost of two points x and y.](image)

In a similar way, we can introduce the connection cost of a point $x$ to a non-empty subset $Y$ of $\mathbb{R}^2$:

$$
\forall x \in \mathbb{R}^2, \quad C_f(x, Y) = \begin{cases} 
\inf \{ \lambda \in \mathbb{R} / \overline{\delta}_{f,\lambda}(x, Y) < +\infty \} & \text{if } x \notin Y \\
-\infty & \text{otherwise.}
\end{cases}
$$

Bronchi location using the connection cost algorithm is realized as follows. In the initialization step, the following images are considered:

• IM1 is the image (binary or labeled) associated with $Y$,
• IM2 is the grey-level image associated with $f$,
• IM3 is defined by: $\text{IM3}(x) = \begin{cases} 
\text{IM2}(x) & \text{if } \text{IM1}(x) \neq 0 \\
+\infty & \text{otherwise.}
\end{cases}$

Secondly, the algorithm performs recursively, back-and-forth and until convergence (i.e., idempotence), the following operation:
if IM1(x) = 0, then IM3(x) ← max(IM2(x), min \( \min_{v \in V(x)} \) IM3(v)),

where V(x) is the elementary upstream neighborhood of x with respect to the current scanning.

The final image is obtained by performing the following operation:

if IM1(x) > 0, then IM3(x) ← -\( \infty \).

This algorithm results in “filling” all the local valleys present in the image. A bronchi marking is obtained by an auto-adapative thresholding of the difference between IM3 and IM2 (Figures 3,4).

![Figure 3: Connection cost-based marking algorithm: principle.](image)

![Figure 4: Connection cost-based marking algorithm: results.](image)

4a: Initial image (IM).
4b: Connection cost with respect to the initial image (CC).
4c: Difference CC-IM.
4d: Marker image obtained from fig. 4c by a self-adaptive thresholding.

The third step of the method concerns the accurate segmentation of the marked object. A contour approach has been adopted. It is based on the concept of **watershed under constraints**\(^\text{17,18} \). In the following paragraph, we shall recall the watershed definition.
Let $f$ be a continuously differentiable function from $\mathbb{R}^n$ to $\mathbb{R}$ with connected compact support, such that the regional minima are the only plateaus. Then, the set of monotonic paths from $x$ to $y$ ($x \neq y$), $\Gamma_{x,y}$, is defined as:

$$\Gamma_{x,y} = \left\{ \gamma \in \partial \text{supp}(f), \quad \forall (s_1, s_2) \in [0,1]^2, \quad s_1 < s_2 \Rightarrow f(\gamma(s_1)) \leq f(\gamma(s_2)) \text{ or } f(\gamma(s_1)) \geq f(\gamma(s_2)) \right\}$$

We will refer to the descending path from $x$ to $y$ as a monotonic decreasing path.

Let $M = (\cup M_h)_{h \in \mathbb{R}}$ be the set of the regional minima of $f$. The catchment basin associated with $M_h$, denoted by $\text{CB}_f(M_h)$, is the interior of the set of points $x \in \mathbb{R}^n$ for which all the descending paths of greatest slope originating from $x$ lead exclusively to $M_h$.

The watershed relative to $M = (\cup M_h)_{h \in \mathbb{R}}$, denoted by $\text{WS}(f)$, is the complement of the union of all Cbs:

$$\text{WS}(f) = \left[ \bigcup_{h \in \mathbb{R}} \text{CB}_f(M_h) \right]^c$$

The catchment basins of the image constitute the attraction zones of its minima.

We notice that the contours of bronchi correspond to the closed ridge lines of the gradient, thus we can segment the image by looking for the watershed of the gradient. As the watershed of the gradient is generally oversegmented, it is convenient to modify the gradient function from which the watershed is calculated (Figure 5). A suitable solution is to use the marking functions for the bronchi and for the background, and to impose the marker set previously found as the gradient minima of the original image before computing the new watershed.

We thus obtain the bronchi contours and we can subsequently evaluate the surface of bronchial sections and their diameters.

![Figure 5: Marking technique based on WS](image)

**2 METHOD VALIDATION**

Once the segmentation method has been developed, it is necessary to evaluate its performances in terms of accuracy for estimating the bronchial caliber. Complementary approaches have been applied in order to test, successively, subjectively, statistically, objectively and mathematically, the accuracy of the measurements.

The first elementary but absolutely essential validation step is according to the degree of satisfaction given by the expert radiologists. In a second step, a statistical validation has been performed over a large representative sampling. Then, an absolute validation using calibrated images, so-called phantoms, has been performed, providing an objective estimation of the method's performances.

We have realized three phantoms. The first consisted in a plexiglas block perforated with calibrated holes of variable dimension and plunged into water. The second was realized with six glass pipettes, perfectly calibrated, planted into a
sponge, all plunged in ultra sound gel. The third was made from calibrated plastic syringes planted into a sponge, all plunged in ultra sound gel.

Finally, a mathematical bronchi modeling-based validation method has been developed. Let us describe in detail the two steps of the modeling. The analysis step makes the extraction of the major bronchi characteristics and parameters possible, and the simulation step makes it possible to synthesize bronchi.

### 2.1 Bronchus analysis

After a study of 80 bronchi from a set of 15 HRCT slices, the following set of characteristics has been selected. The bronchus and its satellite artery are divided into four regions (Figure 6):

1. **the bronchus inner** corresponding to the lumen, with an elliptic shape and a low grey level;
2. **the bronchus wall** whose thickness varies from 1/5 to 1/4 of the bronchus diameter. The grey levels in this area are very different from the bronchus inner grey levels. The grey level values are maximum along the central corona of the bronchus wall and decrease to the boundaries. The maximum value of the bronchus wall depends on the pulmonary environment.
3. **the artery inner** has the same grey level value as the maximum of the bronchus wall.
4. **the artery wall** has the same grey level value as the artery inner, thus it is impossible to distinguish between them. A penetration effect into the surrounding pulmonary tissue can be observed as in the case of the bronchus wall.

**Figure 6:** Bronchus structure.

**Figure 7:** Bronchus profile along the white axis.

As a result of the study of bronchus profiles (Figure 7) it is possible (i) to select the following parameters:

- **grey level parameters:**
  - MaxBWall: the maximum grey level of the bronchus wall,
  - MinBInn: the minimum grey level of the bronchus inner,
  - Env: the grey level of the environment pulmonary tissue.

- **shape parameters:**
  - diameter (expressed in pixels),
  - deformation: it is the ratio between the height and the width of the bronchus including rectangle.

and (ii) to model the bronchus wall by a Gaussian function:

\[
f(x) = \text{MaxBWall} \exp \left( -\frac{1}{2} \left( \frac{x - m}{s} \right)^2 \right),
\]

where \( m \) stands for the point abscissa of maximum grey level and \( s \) is the standard deviation.

The profile of the bronchus inner can be approximated by a Butterworth polynomial:

\[
f(x) = (\text{Max} - \text{Min}) x^n + \text{Min},
\]
where Max (resp. Min) is the maximum (resp. minimum) grey level inside the bronchus and \( n \) represents the profile flatness order inside the bronchus. The value of \( n \) is chosen according to the bronchus diameter. Thus, for bronchi with a diameter greater than or equal to 3 mm, \( n = 6 \), for bronchi of 2 mm diameter, \( n = 4 \), and for bronchi of 1 mm diameter, \( n = 2 \).

NB., as the result of an empirical study, bronchi less than 3 mm in diameter are considered as “small”. Bronchi greater than 3 mm in diameter are called “big”.

A contrast analysis between the maximum level of the bronchial wall and the grey level of the lung parenchyma revealed that the grey levels of each bronchus are adapted to their environment. The contrast definition was inspired from a study on a statistic modeling of mammary micro-califications\(^{19}\):

\[
CO = \frac{\text{MaxBW} - \text{Env}}{\text{MaxBW} + \text{Env}}.
\]

The following results have been obtained on a representative bronchi sampling. If the bronchus diameter is greater than or equal to 3 mm, then

\[
CO = (-5.5 \times 10^{-3}) \text{Env} + 0.835, \tag{4}
\]

if not, the contrast is almost constant and the following value is assigned:

\[
CO = 0.345. \tag{5}
\]

The contrast coefficient makes it possible to estimate the maximum level of the wall when the environment value is known:

\[
\text{MaxBW} = \text{Env} \frac{1 + \text{CO}}{1 - \text{CO}}. \tag{6}
\]

### 2.2 Bronchus synthesis

Taking into account the parameters defined in the analysis step, a complete model of bronchus together with its associated artery has been proposed. Firstly, the diameter and the position of the bronchus are chosen by the user. Then, the remaining parameters are set as described in the following paragraphs.

The bronchus wall (resp. the artery wall) thickness is taken between 1/5 and 1/4 of the bronchus diameter. For a bronchus diameter less than 2 mm, the thickness coefficient is set to 1/4 in order to have at least one pixel for the wall. Also, in this case, the probability of having an associated artery is 50%.

To simulate bronchi having an oblique orientation with respect to the slice level, cross sections of simulated bronchi are elliptic in shape. The ratio between the major and the minor ellipsis axis is called the flatness coefficient (denoted by \( c \) in the next paragraphs) and it is randomly chosen between 1.1 and 1.43 for “big” bronchi, and between 1.1 and 1.25 for “small” bronchi. The cross section orientation of the simulated bronchi is also randomly chosen, the rotation angle being drawn from the interval \([-\pi, \pi]\), with a \( \pi/16 \) step.

The pulmonary artery homologous to the bronchus, if present, is generated with the same size, shape and orientation as the bronchus.

The contrast coefficient is calculated according to the bronchus size and the average grey level of the surrounding lung parenchyma. If the average grey level of the pulmonary environment is into the interval \([70, 90]\), the formulas (4) and (5) are used to determine the maximum grey level of the bronchial wall. Otherwise, MaxBW is set to 185 for “big” bronchi and to 160 for the “small” bronchi. Similarly, the minimum grey level into the bronchial lumen (MinBlnm) is defined to be 0 for “big” bronchi and 8 for the “small” bronchi.

The symmetry property of the bronchus is taken into account for calculating the grey level distribution by using polar coordinates. As mentioned in the analysis step, the bronchus internal profile is modeled by a Butterworth polynomial centered at the origin whereas the bronchus (resp. artery) wall is modeled by a Gaussian function, and the artery inner with a
constant. In this case, the parameter \( x \) from the formulas (1) and (2) stands for the radius in polar coordinates (while the angle varies from 0 to \( 2\pi \)).

The last step of the synthesis consists in edge smoothing in order to simulate the interpenetration effect between the bronchial wall and the lung parenchyma. This makes the realism of the final image possible (Figures 8,9). The smoothing operator is an 8-connectivity average filter acting over an area which includes the bronchus external half wall and a band belonging to the pulmonary parenchyma (Figure 10).

\[ S_{\text{theoretical}} = \pi \frac{d^2}{4} c, \]

where \( c \) stands for the bronchus flatness coefficient and \( d \) for the bronchus diameter (in millimeters).

2.3 Designing criteria for accuracy estimation

Once the bronchus model is achieved and the simulated bronchi are generated, a segmentation using the algorithm presented in Section 1 is applied. To establish segmentation accuracy, the surface measurement error is studied. Three types of surfaces are considered:

- the theoretical surface issued from the continuous model of the synthesized bronchus:

\[ S_{\text{theoretical}} = \pi \frac{d^2}{4} c, \] (7)

- the surface obtained from the continuous bronchus.

- the surface obtained from the simulated bronchus.

Figure 8: Simulated bronchus profile along the white axis:
8a: Without edge smoothing,
8b: With edge smoothing.

Figure 9: 3D meshes and their corresponding images:
9a: Without edge smoothing,
9b: With edge smoothing.

Figure 10: Smoothing zone.
• the discrete surface issued from the discretization of the theoretical surface imposed by the finite resolution of the image. It is evaluated simultaneously with the bronchus simulation.

• the measured surface, which is the surface of the segmented bronchus. For square frame images, the surface measurement verifies the following equation:

\[
S_{\text{measured or discrete}} = \text{Nb\_pixels} \left( \frac{\text{Field}}{\text{Resolution}} \right)^2,
\]

where:

\( \text{Nb\_pixels} \) is the number of pixels of the mask,
\( \text{Resolution} \) is the image dimension in pixels (here 512),
\( \text{Field} \) represents the field of view for the HRCT reconstructed images (here 131 or 262 mm).

The measured accuracy is expressed in terms of relative errors between the measured surface as compared to the theoretical and discrete surfaces and, respectively, between the estimated diameter versus the theoretical diameter:

\[
\varepsilon_{\text{rel}} = \frac{\Delta S}{S}, \quad \varepsilon_{\text{rel}} = \frac{\Delta d}{d}.
\]

3 RESULTS AND DISCUSSION

This section presents the results of the validation step (Section 2) applied to the segmentation method (Section 1).

The method was validated at first, due to the degree of satisfaction as attested to by expert radiologists as a result of its high performance. Nevertheless, in a second phase a greater interactivity was achieved, making it possible for the physicians to remove a segmented bronchus (typically when it is too oblique) or to add a bronchial section missed in the first segmentation step. After these options were added, the expert radiologists claimed to be completely satisfied.

The statistical validation experimentally proved the versatility of the segmentation method with respect to the signal heterogeneity and that it is completely independent of the bronchi location.

The validation on phantoms proved the pertinence of the segmentation method and the high level of measurement accuracy for "big" bronchus caliber. Nevertheless, this step underlined the difficulties to obtain an objective validation because of the phantom imperfection and led to the development of mathematical simulation based on bronchus modeling. The results from the mathematical validation are discussed in the following paragraphs.

A non-negligible source of errors is represented by the surface discretization. Comparisons between the measured surface and the theoretical surface are affected by this kind of error. Figure 11 presents the distribution of the discretization error with respect to the theoretical surface, where

\[
\text{Discretization Error} = \frac{\text{Pixel Surface}}{\text{Bronchus’ Theoretical Surface}} \times 100 \% \quad \text{or} \quad \left( \frac{\text{Field}}{\text{Resolution}} \right)^2 \times \frac{\text{100 \%}}{\text{Bronchus’ Theoretical Surface}}\%.
\]

It is shown that, for a 131 mm field, missing one pixel introduces an 8.3\% error for a 1 mm bronchus diameter whereas the error is only 0.08\% for a 10 mm bronchus diameter. Similarly, for a 262 mm field, the error is 33.2\% for a 1 mm bronchus diameter and 0.33\% for a 10 mm bronchus diameter.

Measures have been realized over a total of 30 sheep lung images (15 images of 131 mm field and 15 images of 262 mm field). The 131 mm field images have been computed by the scanner from the corresponding 262 mm field images. Among these fifteen image pairs, nine are acquired in a natural state, three after the administration of histamine, one after a
histamine and nitrogen monoxide treatment, and the last two following a nitrogen monoxide injection. Six hundred synthesized bronchi have been segmented from this set of images. Three main parameters have been varied during this study, the bronchus diameter, from 1 mm to 10 mm, the flatness coefficient chosen between 1, 1.2 and 1.4, and the rotation angle between 0, π/6 and π/4.

Figure 12 shows the surface (resp. diameter) measurement relative errors with respect to the bronchus diameter both for the 131 mm field images and the 262 mm field images. The surface relative error was estimated according to the theoretical bronchus surface:

$$e_{th. rel} = \frac{\text{Measured Surface}}{\text{Theoretical Surface}} - 1 \times 100 \{\%\} . \quad (11)$$

Similar results are obtained for the relative error estimated with respect to the discrete surface:

$$e_{dis. rel} = \frac{\text{Measured Surface}}{\text{Discrete Surface}} - 1 \times 100 \{\%\} . \quad (12)$$

The segmentation was successfully achieved for all simulated bronchi in 131 mm field images. For the 262 mm field images, all synthesized bronchi were successfully segmented except bronchi having a 1 mm diameter. However, this behaviour was expected since a 262 mm field corresponds to a 0.512 mm pixel width, meaning that there are not enough pixels to represent the bronchus and the wall correctly.

The measure accuracy can be qualified as good as the bronchus diameter is large enough so that the errors due to the discretization do not have a significant influence. Thus, for 131 mm field images the surface estimation can be considered
reliable for bronchi with diameters between 2 and 10 mm. For 262 mm field images, the same behaviour is achieved for bronchi with diameters between 4 and 10 mm.

The segmentation and measurement functionalities are available by using a popup menu integrated in a user-friendly interface designed in C++ under the Microsoft Windows 3.1 environment. The software package which has been developed provides a non-invasive tool for an accurate and reproducible quantitative analysis of bronchial caliber in HRCT images.

**CONCLUSION**

We have presented an accurate and a robust method for a quantitative analysis of bronchial caliber in HRCT images. This method combines morphological filtering, connection cost-based marking, and conditional watershed techniques.

The method has been completely validated on simulated bronchi resulting from a mathematical modeling. For images in 131 mm field of view, the segmentation was successfully achieved whatever the bronchus diameter. In this case, the measurement accuracy for bronchi having a diameter greater than 2 mm was 95.3%, 90.9% for bronchi having a 2 mm diameter, and 76.8% for bronchi having a 1 mm diameter. For images in 262 mm field of view, only the 1 mm diameter bronchi were not segmented. The accuracy for bronchi having a diameter greater than 3 mm was 91.7%, and 76.8% for bronchi having a diameter of 2 and 3 mm.

Our future objective consists in developing a 3D lung reconstruction technique which will make it possible to provide a more precise analysis of the pulmonary function. In particular, the bronchial caliber will be estimated from slices reconstructed in a plane perpendicular to the bronchus axis.

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