



POLITECNICO
MILANO 1863

SCUOLA DI INGEGNERIA INDUSTRIALE
E DELL'INFORMAZIONE

A reliable sinonasal CT segmentation procedure for the CFD of the nasal airflow

TESI DI LAUREA MAGISTRALE IN
AERONAUTICAL ENGINEERING - INGEGNERIA AERONAUTICA

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Academic Year: 2024-2025

Abstract

This thesis is part of a collaboration between Politecnico di Milano and the Otolaryngology Department of A.O. San Paolo in Milan, aimed at improving diagnostic tools through Computational Fluid Dynamics. Part of the project relies on a Machine Learning approach to identify flow patterns associated with specific nasal pathologies. This requires reliable CFD simulations of patient-specific nasal airways. The main objective of this work is to develop a reliable and repeatable procedure for reconstructing three-dimensional geometries of the nasal cavity from CT scans. Particular attention is given to the treatment of accessory cavities connected to the main nasal airway (such as the paranasal sinuses), whose removal significantly simplifies the computational geometry. High-fidelity, time-resolved CFD simulations (DNS and LES) are performed to assess the impact of sinus removal from both a global and a local perspective. Results show that removing the paranasal sinuses does not significantly affect the global nasal flow rate or the nasal resistance. Differences are found in localized regions in proximity to the closed ostia, where small variations in mean velocity and fluctuation intensity can be observed. In particular, it is discovered that oscillations tend to decrease after sinus removal. Overall, the proposed simplification preserves the functional fidelity of the model, enabling an efficient and standardized reconstruction workflow suitable for CFD simulations.

Keywords: CFD, DNS, flow inside the nasal cavities, CT scan segmentation

Abstract in lingua italiana

Questa tesi è parte di una collaborazione tra il Politecnico di Milano e il Dipartimento di Otorinolaringoiatria dell'Ospedale San Paolo di Milano, finalizzata al miglioramento degli strumenti diagnostici tramite l'utilizzo della Fluidodinamica Computazionale (CFD). Una parte dell'attività prevede l'utilizzo di algoritmi di Machine Learning per identificare pattern di flusso associati a specifiche patologie nasali. Ciò richiede la realizzazione di simulazioni CFD accurate all'interno delle cavità nasali di pazienti reali. L'obiettivo principale di questo lavoro è lo sviluppo di una procedura affidabile e ripetibile per la ricostruzione di geometrie tridimensionali delle cavità nasali a partire da scansioni TC. Particolare attenzione è dedicata al trattamento delle cavità accessorie connesse ai canali nasali principali (i seni paranasali, per esempio) la cui rimozione consente una significativa semplificazione della geometria computazionale. Per valutare l'impatto della rimozione dei seni, sia a livello globale sia locale, sono state eseguite simulazioni CFD ad alta fedeltà, risolte nel tempo (DNS e LES). I risultati mostrano che la rimozione dei seni paranasali non altera in modo significativo né il flusso globale né la resistenza nasale. Le differenze riscontrate risultano limitate a regioni localizzate in prossimità degli ostii chiusi, dove si osservano piccole variazioni sia nella velocità media sia nell'intensità delle oscillazioni. In particolare, si è notato che le fluttuazioni tendono a ridursi in assenza dei seni. Nel complesso, la semplificazione proposta preserva la fedeltà funzionale del modello, consentendo una procedura di ricostruzione efficiente e standardizzata, idonea alla generazione di modelli per simulazioni CFD.

Parole chiave: CFD, DNS, flusso nelle cavità nasali, ricostruzione da scansioni TC

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Introduction

Why it is important to study the nose flow

The nose is much more than two cavities behind a projection at the center of the face. Although its primary function is to channel air to the lungs for respiration, the nasal apparatus also humidifies, filters, warms, and senses the air we breathe [20]. A healthy nasal airway is therefore essential for overall wellness during everyday life and during physical activity. Efficient nasal breathing ensures the right amount of oxygen to the body, starting with the brain and other vital organs.

In spite of its importance, bad nasal breathing is extremely common. Studies report that between 50% and 60% of the population has some degree of nasal obstruction, making nasal breathing difficult [6, 37] resulting in inability to sleep well at night, somnolence, and poor concentration [21]. Among these patients, around 76% present a form of septal deviation, while 72% show inferior turbinate hypertrophy [6].

For the most severe cases, surgical interventions such as septoplasty and turbinate reduction are available. However, their outcome is generally unsure. These surgeries are complex and often fail to resolve the patient's symptoms, with a large fraction of patients reporting little or no improvement after surgery. In particular, dissatisfaction rates following septal deviation correction range between 50% and 80% [17, 44]. In some cases, surgery may even lead to new symptoms, such as empty nose syndrome [3, 27].

Such a low success rate of nasal surgeries is largely due to the extremely complex geometry of the nasal airways. This complexity strongly influences nasal functionality. However, the relationship between anatomical shape and function is still not fully understood [10]. Furthermore, there is a big inter-patient anatomical variability, and otolaryngologists are often required to make surgical decisions based primarily on visual inspection and experience. It is well known that unusual anatomies can be completely asymptomatic, while apparently "normal" nasal geometries may lead to breathing difficulties. As a result, defining a functionally average healthy nasal anatomy remains an unsolved problem [5].

In an effort to provide support to doctors dealing with nasal obstruction, the OpenNOSE

project was established as a collaborative effort between the Otolaryngology Department of A.O. San Paolo in Milan and the Department of Aerospace Science and Technology at Politecnico di Milano.

This thesis is part of the OpenNOSE project and continues the work already done within this framework. The main objective of the project is to develop an open-source diagnostic tool capable of supporting clinical decisions by reducing uncertainty and complementing traditional visual analysis with computational information. To reach this goal, the project employs engineering approaches, such as Computational Fluid Dynamics (CFD), to quantitatively assess nasal airflow and its relationship with anatomical features.

Thesis objectives

This thesis is positioned within the broader goal of revisiting the entire OpenNose workflow, which starts from nasal CT scans and the reconstruction of 3D nasal geometries, proceeds through the construction of a synthetic training database for a Machine Learning model and finally aims at extracting diagnostic information (see Section 1.3 for a detailed description).

In particular, this work focuses on the initial stages of the pipeline: the **segmentation procedure** used to reconstruct three-dimensional geometries of the nasal airways from CT scans, and the manual **creation of anatomical defects** required to build the synthetic training dataset. However, only the reconstruction workflow is discussed in this document, whereas the defect-generation process, which has absorbed a substantial effort and time, is reported separately in Appendix A.

This document therefore describes the development of a reliable and repeatable procedure for generating computational geometries of the nasal airways starting from CT data. The resulting 3D models are intended to be a simplified version of the nasal cavity, in which those auxiliary anatomical structures which are assumed to not alter significantly the overall airflow are removed. However, such simplifications must not compromise the fluid-dynamic accuracy nor affect substantially the functional behavior of the nose.

Particular attention is dedicated to the representation of the ostia, i.e., the openings connecting the main nasal cavity to the paranasal sinuses (see Chapter 1). An in-depth investigation is conducted to assess the impact of removing these structures on the airflow, both at a global and at a local scale.

The key question addressed by this second objective is:

How much does the nasal airflow change when the sinuses are removed? Does the presence of a second ostium modify this behaviour?

As shown in previous studies [40, 52], the airflow inside the paranasal sinuses is extremely low and generally negligible when compared to the main nasal airflow. Therefore, a reasonable assumption is that removing the sinuses will not significantly affect the macroscopic quantities of nasal airflow. In particular, global quantities such as total flow rate, nasal resistance, and averaged flow quantities observed from "afar" are not expected to vary in a significant or appreciable way among the three configurations.

On the other hand, when focusing on those regions where the geometry has been changed, local flow variations are expected, at least in terms of velocity magnitude. These differences are expected to be localized around of the modified regions. A first way to assess these variations is by analyzing the velocity magnitude U_{mag} , in order to quantify their intensity.

Moreover, additional differences are expected in the geometry characterized by the accessory maxillary ostium. In a physiologically normal configuration, the maxillary sinus is connected to the nasal cavity through a single hole, implying that all the air entering the sinus must exit through the same opening, resulting in a zero net mass flow. In contrast, the presence of two ostia effectively introduces an additional flow path. As a consequence, a non-zero net flow through the maxillary sinus is expected.

To verify that the proposed reconstruction procedure does not compromise the relevant fluid-dynamic characteristics, a series of comparative Computational Fluid Dynamics (CFD) simulations were performed. Both Direct Numerical Simulation (DNS) and Large Eddy Simulation (LES) approaches were employed.

The segmentation workflow is based on two open-source software: *3D Slicer* and *Blender*. All CFD simulations are carried out using *AMPHIBIOUS*, a new DNS/LES solver developed at the Department of Aerospace Science and Technology of Politecnico di Milano. Since this thesis represents the first applications of *AMPHIBIOUS* to a problem of this kind, a brief description of the solver and the motivations behind its choice will be provided.

Finally, all the work related to anatomical accuracy were reviewed by an otolaryngologist ensuring consistency and fidelity with respect to the original patient anatomy.

Thesis structure

- **Chapter 1: Preliminary concepts** Introduction to nasal anatomy, pathologies, and CT scans, providing the background knowledge required to fully understand this work.
- **Chapter 2: Segmentation procedure** Detailed description of the reconstruction procedure used to generate computational surfaces from CT scans.
- **Chapter 3: CFD simulations** Description of the geometries, simulation setup and software employed.
- **Chapter 4: Results** Analysis and comparison of similarities and differences observed in the simulated airflow.
- **Appendix A: Defects manufacturing** In-depth description of the procedures used to introduce anatomical pathologies into healthy geometries.
- **Appendix B: Additional images** Collection of supplementary simulation images used to complement the results presented in the main body of the thesis.

1 | Preliminary concepts

1.1. Nose anatomy and physiology

Since most of this work focuses on the internal geometry of the nose (especially the nasal sinuses removal and the addition of defects), a brief overview of the anatomy of the nasal cavity is provided here. For a more detailed description, the reader is referred to subject-specific anatomical textbooks such as [20] and [51].

The nose is a complex anatomical structure composed of bone, cartilage, and soft tissues. It is conventionally divided into two main parts: the external nose and the internal nose.

The external nose is the portion that protrudes from the face and has an approximately pyramidal shape. Since its external shape does not affect the airflow within the nasal cavities, its detailed description is not required for this work.

The internal nose, or nasal cavity, is a paired structure composed of two chambers (nasal cavities or fossae) separated by the nasal septum. It extends from the nostrils to the nasopharynx and has an average length of 6–7 cm. The nasal cavity runs above the oral cavity and is separated from it by the palate. Posteriorly, it is connected to the pharynx through the nasopharynx, which connects the nasal passages to the rest of the upper airway.

The internal nose is mainly composed of the nasal septum, the turbinates (or nasal conchae), and the paranasal sinuses.

The nasal septum is a thin vertical structure made of cartilage anteriorly and bone posteriorly. It separates the nasal cavity into left and right passages.

The turbinates, also known as nasal conchae, are curved, shelf-like bony structures that protrude from the lateral walls of the nasal cavity. They are covered by mucosa and have a characteristic shell-like shape. The space beneath each turbinate is referred to as a meatus.

In humans, there are typically three turbinates on each side: inferior, middle, and superior.

The inferior turbinate is the largest of the three, it extends along the lower part of the nasal cavity and defines the inferior meatus. The middle turbinate occupies a central position. The superior turbinate is the smallest and is located high in the nasal cavity, close to the olfactory region.

Together, the turbinates give the nasal cavity its characteristic hook shape. Their presence significantly increases the internal surface of the nasal cavities, enhancing thermal exchange, humidification, and filtration of the air. Figure 1.1 provides a clear visualization of the nasal cavity and its internal structures, shown through a drawing and a corresponding CT scan of the same anatomical region. Another CT scan section can be seen in Fig. 1.2 showing a coronal section 1.2(a) taken deeper within the nose geometry.

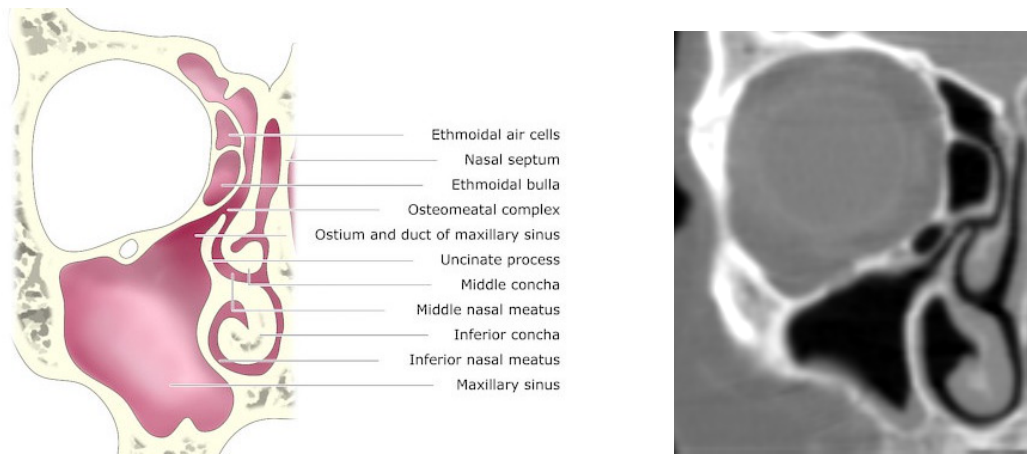


Figure 1.1: Nasal cavities and its structure. Coronal section showing the maxillary sinus and its connection to the middle meatus. Drawing and CT scan of the same region. Left image from [13].

The paranasal sinuses are air-filled cavities located within the bones of the skull and surrounding the nasal cavity. They are lined with mucosa and communicate with the nasal passages through small openings called **ostia**. The paranasal sinuses contribute to voice resonance, reduce the weight of the skull, and may play a role in conditioning inhaled air, although their exact functional contribution to nasal airflow remains partially debated [20].

There are four major nasal sinuses: the frontal, maxillary, ethmoidal, and sphenoidal sinuses.

The frontal sinuses are located within the frontal bone above the orbits. They drain into the anterior portion of the middle meatus through the frontal recess.

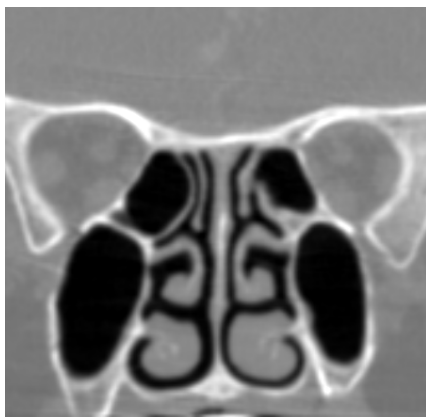
The maxillary sinuses are the largest of the paranasal sinuses and are located within the

maxillary bones (cheekbones). They drain into the middle meatus through the ostiomeatal complex (via the infundibulum), a key anatomical region located above the middle meatus. The drainage path is partially bounded by the uncinate process, a thin, sickle-shaped bony projection of the ethmoid bone.

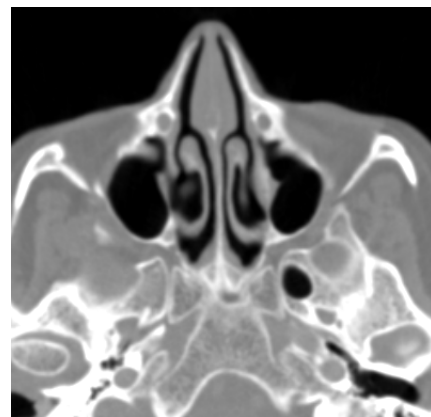
The ethmoidal sinuses, also known as the ethmoidal labyrinth, consist of multiple small air-filled cavities called ethmoidal air cells, located between the nasal cavity and the orbits. The anterior and middle ethmoidal cells drain into the middle meatus, while the posterior ethmoidal cells drain into the superior meatus.

The sphenoidal sinuses are located deep within the skull, posterior to the nasal cavity. They open into the sphenoethmoidal recess, a small space situated above and posterior to the superior turbinate.

An additional anatomical structure is the nasolacrimal duct, which drains tears from the eyes into the nasal cavity. It opens into the inferior meatus.



(a) Coronal section



(b) Axial section

Figure 1.2: CT scans showing coronal and axial sections of the same healthy patient. In the left image, the three nasal turbinates are clearly visible, together with the maxillary and sphenoidal sinuses. The right image shows an axial view highlighting the inferior concha and the maxillary sinuses. The openings of the naso-lacrimal ducts are also visible.

1.2. Pathologies

As already mentioned in the introduction, a large fraction of the population, almost one out of three people, suffers from some sort of nasal disorder. Reporting here an exhaustive list of all possible nasal pathologies would be redundant and unnecessary, as only a limited subset is relevant for this work. For a broader overview of nasal pathologies, the reader

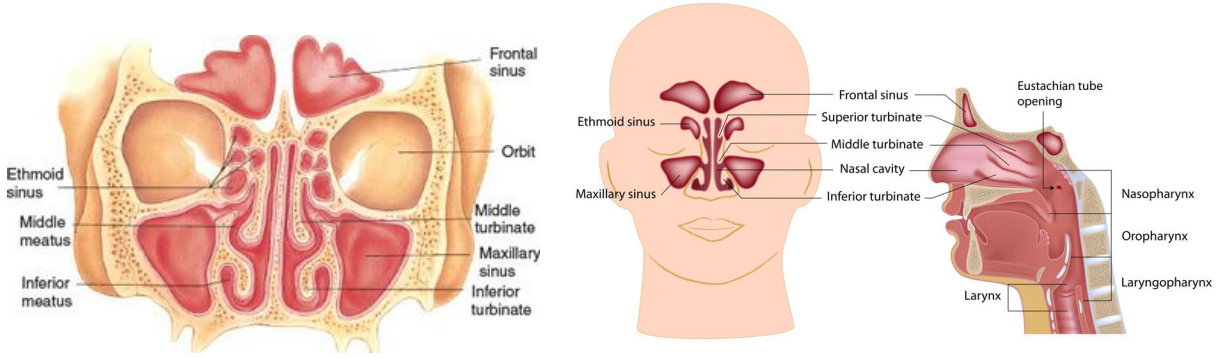


Figure 1.3: Schematic representation of the paranasal sinuses in both a coronal and a sagittal section. Images from [4].

is referred to more in-depths reviews such as [42].

Within the OpenNOSE framework, a limited but representative selection of nasal pathologies is considered. This choice does not imply a loss in generality, since these conditions account for the majority of cases [6] and are commonly treated via surgery. The pathological anatomical variants considered in this work are divided into two main categories: turbinate hypertrophy and septal variations. The second category includes both septal deviations and septal crests [46].

Turbinate hypertrophy is defined as an enlargement of a turbinate that may involve the bone and the mucosa, or both [21, 22]. This enlargement leads to a reduction of the nasal cross-sectional area and consequently to nasal obstruction and difficulty in breathing. A CT scan of a patient affected by left inferior turbinate hypertrophy [19] is shown in Fig. 1.4(a), together with a non-hypertrophic case shown for comparison in Fig. 1.4(b).

Septal deviation is the most common anatomical cause of nasal obstruction [6, 22]. A deviated septum is characterized by a displacement of the nasal septum from the midline, resulting in an asymmetric nasal cavity. Based on the shape and severity of the deviation, septal deviations can be classified as linear (I-shaped), C-shaped, or S-shaped [30]. As a consequence of this asymmetry, one nasal cavity becomes smaller, leading to breathing difficulty. Figure 1.5 shows examples of septal deviations with different morphologies and severity.

Septal crests are bony protuberances that extend from the nasal septum [46]. These structures can significantly modify the internal nasal geometry, contributing to breathing difficulty and discomfort. Septal crests may occur at different locations along the septum. Figure 1.6 illustrates two common configurations: a basal crest, which is frequently observed, and a lateral crest.

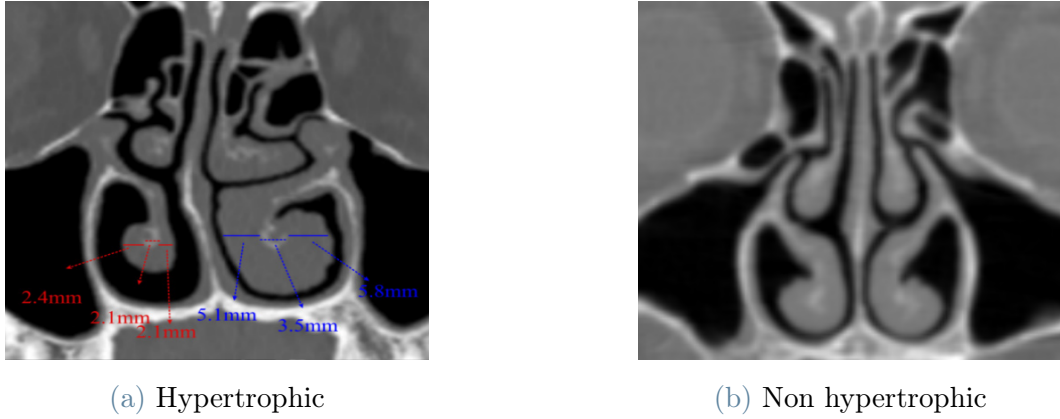


Figure 1.4: CT scans showing inferior turbinate hypertrophy and a normal anatomy. The left image (a) is a coronal section of a patient with left inferior turbinate hypertrophy. The right image (b) is a coronal section of a non-hypertrophic subject shown for comparison. The patient in (a) also presents a mild septal deviation to the right. Left image from [19].

Additional information can be found in Appendix A.

1.3. Nose surgeries and the OpenNOSE project

As already mentioned in the introduction, the success rate of nasal surgery is quite low, with more than one out of two patients being dissatisfied with the surgical outcome [17, 44]. Surgeons are required to operate within an intrinsically complex geometry, with a lot of inter-patient anatomical variability, and must generally rely on experience and visual assessment. Since the relationship between nasal shape and function is highly non-linear (the connection being the Navier-Stokes equations) it is extremely challenging to predict or achieve the desired outcome through surgery.

At present, the only available tool providing a numerical parameter directly related to nasal fluid dynamics is *rhinomanometry* [49]. Rhinomanometry is a non-invasive examination that provides an objective measurement of the *nasal resistance*, defined as the ratio between the pressure drop ΔP across the nasal airway and the nasal airflow rate Q :

$$R = \frac{\Delta P}{Q}.$$

In practice, symptoms of nasal obstruction are commonly assessed using subjective questionnaires such as the Nasal Obstruction Symptom Evaluation (N.O.S.E.) score [18, 50]. However, it is well established that nasal obstruction symptoms are not always directly correlated with a high nasal resistance [29, 36]. A clear example is the *empty nose syn-*



(a) Patient 1, coronal section



(b) Patient 2, axial section

Figure 1.5: CT scans showing septal deviations in two different patients. Both patients present septal deviation to the right. Patient 1 (a) presents a severe septal deviation due to the large concha bullosa of the left middle turbinate. Patient 2 (b) presents a less pronounced septal deviation.

drome [16], a pathological condition that may arise after surgical procedures aimed to reducing nasal resistance (e.g. turbinectomy), where patients experience obstruction symptoms despite a reduced aerodynamic resistance.

In addition, the reliability of rhinomanometry as a quantitative measure of nasal resistance is being debated. Recent studies report significant incongruencies between numerically computed nasal resistance and clinically measured values. In some cases, the mismatch can exceed 100% [2].

In this context, Computational Fluid Dynamics (CFD) represents an interesting complementary diagnostic tool for otorhinolaryngologists. Over the past decade, numerous studies have employed CFD to reconstruct a complete, patient-specific, nasal airflow starting from CT scans. On the other hand, interpreting CFD results in a clinically meaningful way is hard, primarily because it is impossible to compare patient-specific simulations with those of an *average* healthy nose [5].

In this scenario, Machine Learning (ML) emerges as a powerful tool. ML methods can process large sets of features obtained through CFD (which may not be meaningful alone) and relate them to known pathologies, their severity, and their location, giving relevant information to surgeons and doctors.

One of the goals of the Open NOSE project is precisely this objective: developing an integrated CFD-ML tool capable of transforming simulation outputs into diagnostic in-



(a) Patient 1, coronal section



(b) Patient 2, axial section

Figure 1.6: CT scans of septal deviations due to septal crest for two different patients. Patient 1 presents a deviated maxillary crest, patient 2 presents a lateral septal crests.

formation for surgeons.

It is well known that training such a model requires a very large and labeled dataset. However, such dataset does not exist in the present day, since its construction would require an enormous effort in terms of data acquisition and manual labeling by expert physicians.

Previous works within the OpenNOSE framework developed a procedure capable of generating a synthetic, but anatomically and physiologically accurate, dataset [41] to solve this problem. The workflow is briefly described below, and an illustrative scheme is shown in Fig. 1.7.

Geometry The geometry of a single healthy patient is reconstructed from CT scans, and specific anatomical defects (such as septal deviations or turbinate hypertrophies of various kinds) are manually added. This process generates a family of nasal geometries, each characterized by a unique pathology. Combining different defects, a large number of anatomical variations can be produced, effectively creating a "vector basis" for nasal geometries.

Functional mapping Through a procedure known as functional mapping [26, 32], the manually generated defects are projected onto the geometries of other healthy patients. This step enables an enrichment of the dataset with a very low cost [41].

CFD simulations CFD simulations (from DNS and LES to more recent reduced-order models [11]) are performed on the generated geometries. From these simulations, a set of

numerical features are extracted.

Model training A machine Learning model, a neural network, is trained [8, 41] to classify nasal geometries as healthy or pathological and, in the second case, to identify the type of pathology, its severity, and its anatomical location.

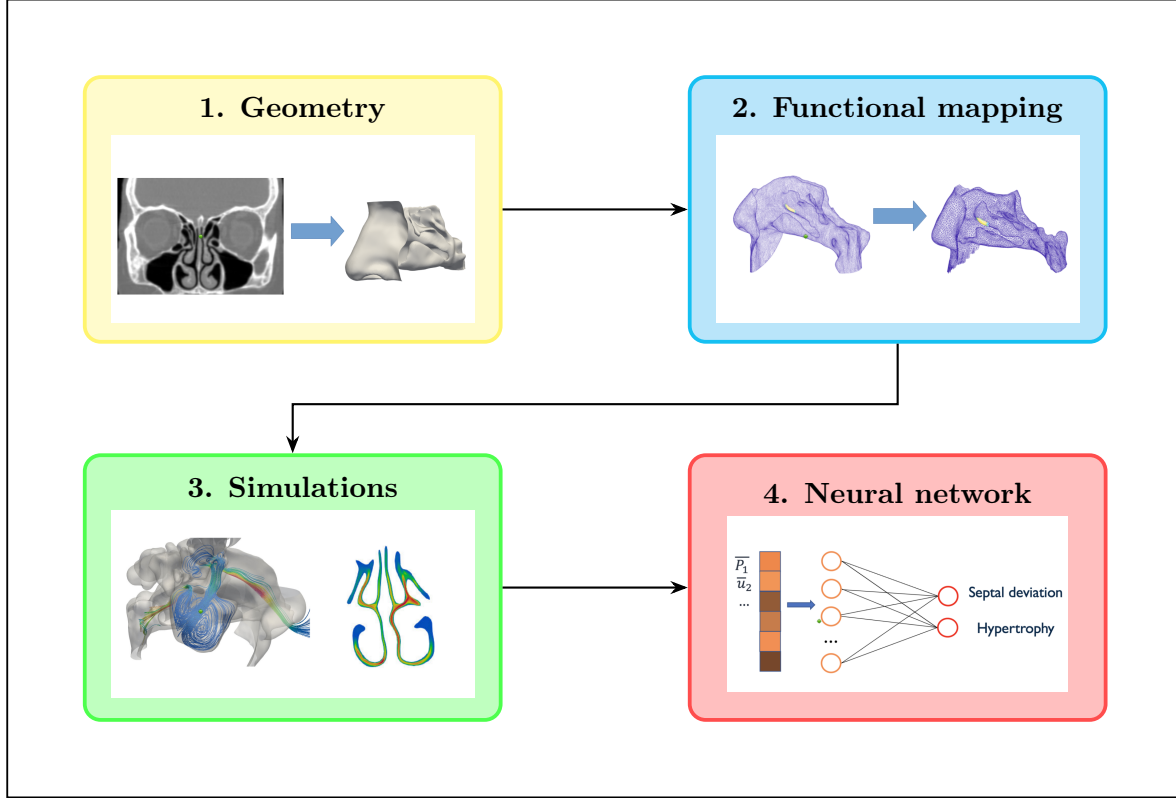


Figure 1.7: Schematic representation of the OpenNOSE pipeline.

The primary objective of this thesis is to address the first step of this workflow by developing a reliable and repeatable reconstruction procedure for nasal geometries. A secondary, but equally important, objective is to enrich the existing dataset by creating those anatomical defects that were not modelled in the original work [41]. This part of the work is described in detailed in Appendix A.

1.4. Computed Tomography

Since a significant part of this work is based on reconstructing 3D geometries starting from CT scans, it is important to briefly describe this imaging technique.

Computed tomography (CT) is a medical imaging technique that is able to producing detailed cross-sectional images (slices) of the human body by combining a rotating X-

ray emitter with computer post-processing. As X-rays pass through the body, they are attenuated by tissues with different physical properties. This attenuation is then measured by detectors positioned on the opposite side from the X-ray source.

The ability of a material to attenuate X-rays is called *radiodensity*. The measured attenuation values are saved and displayed in a 2D image, where each pixel corresponds to a small volume of tissue, called "voxel" in 3D. Each voxel is assigned a grayscale value proportional to its radiodensity: tissues that attenuate more, such as bone, appear bright and white, while weakly attenuating materials, such as air, appear dark.

After having acquiring a large number of consecutive slices, the CT scanner allows to reconstruct a three-dimensional representation of the scanned anatomy. This 3D image enables physicians and experts to visualize internal anatomical structures.

Radiodensity in CT images is quantitatively expressed using *Hounsfield units* (HU). The Hounsfield scale is a linear transformation of the measured linear attenuation coefficient, normalized with respect to the attenuation of distilled water and air under standard conditions. For a voxel with linear attenuation coefficient μ_x , the corresponding Hounsfield value is defined as:

$$HU = 1000 \times \frac{\mu_x - \mu_{\text{water}}}{\mu_{\text{water}} - \mu_{\text{air}}}$$

By convention, water is assigned a value of 0 HU and air a value of -1000 HU.

2 | The segmentation procedure

2.1. Objectives

The primary requirement of the Open Nose project is the definition of a reliable and repeatable procedure capable of generating a usable computational mesh starting from a CT scan. Although this task may appear straightforward, since most DICOM visualization software can automatically produce three-dimensional reconstructions, the complex anatomy of the nasal cavity, combined with the small dimensions of its internal airways, makes the generation of an anatomically accurate model particularly challenging.

As already mentioned, to ensure that the reconstruction procedure yields realistic geometries consistent with the original CT data, the entire workflow and all the results were reviewed step by step.

The reconstruction process is divided into two main stages, each performed using a different software tool: 3D Slicer and Blender. The majority of the segmentation and reconstruction work is carried out in 3D Slicer, while Blender is used for geometry refinement and final cleanup.

The main objective of the procedure is to obtain a clean and simplified representation of the internal nasal airways, excluding the paranasal sinuses and other auxiliary anatomical structures. These elements significantly increase geometric complexity, making the subsequent functional mapping step more difficult, while having a negligible impact on nasal airflow. This assumption will be quantitatively assessed and discussed in Chapter 4.

2.2. 3D Slicer

The first stage of the reconstruction workflow requires the use of 3D Slicer [12], an open-source software platform for the visualization, processing, and analysis of medical and other three-dimensional images and meshes.

A high-quality CT scan is required, with a spatial resolution of at least 0.5 mm. This

requirement arises from the small dimensions of the nasal airways, which are typically only a few millimeters wide and may be significantly narrower in the presence of pathological conditions such as turbinate hypertrophy. Higher image resolution results in a larger number of voxels available for segmentation, allowing for a more accurate reconstruction of the internal geometry [35].

In addition to spatial resolution, the CT scan must be free of artifacts in the region of interest and must fully capture the nasal geometry. In particular, both the superior and inferior extents of the nasal cavity must be included.

The complete procedure described in the following section allows to start from the CT scan data and end up with an exported triangulated surface mesh in STL format.

Reconstruction procedure

1. Load the DICOM dataset.
2. **Crop Volumes module**
 - *Input ROI (Region of Interest)*: create a new ROI.
 - Reduce the ROI around the region of interest, carefully avoiding the removal of any nasal-related cavities.
 - Leave the *Spacing scale* set to 1.00. *
 - Enable *Isotropic spacing*.
 - Select the *B-spline* interpolator.*
3. **Segment Editor module**
 - Add a new segment.
 - Select the *Threshold* tool and set the lower limit to $HU = -300$.**
 - Select the *Logical Operators* tool and apply the *Invert* operation.
 - Select the *Islands* tool and enable *Keep largest island*.
 - Iteratively use the *Paint*, *Erase*, and *Keep largest island* tools to manually remove the paranasal sinuses and the cavities of the ethmoidal labyrinth. This manual step is time-consuming and requires practice as well as a good understanding of nasal anatomy. Care must be taken to remove only the cavities while preserving the connecting airways, in order to remain as faithful as possible to the real geometry. See the figure below.
 - Inspect all views and remove any isolated voxels, as they would generate sharp spikes in the surface mesh. At least four connected voxels should be preserved whenever possible.
 - In the *Show 3D* drop-down menu, set the *Smoothing factor* to 0.8. *
 - Export the surface mesh in STL format using a *Scale* factor of 0.001.

Box 2.1: Reconstruction procedure showing the sequence of required operations. Quantities marked with an asterisk (*) can be customized according to specific needs. Additional care must be used for the choice of the HU threshold (**).

2.2.1. The HU threshold

It is now necessary to provide a detailed explanation of some key parameters and methodological choices adopted in the reconstruction procedure.

The first, and arguably most critical, parameter is the selection of the radiodensity threshold.

The approach described in this work is based on a fixed Hounsfield Unit (HU) threshold to define the general boundaries of the reconstructed model. However, this method is not perfect and may introduce errors or inaccuracies if applied without supervision. For this reason, manual corrections are systematically performed after the initial thresholding step. Alternative approaches have been proposed, for example, machine learning-based methods employ adaptive or data-driven strategies to define the nasal boundaries (e.g., [33, 39]). Nevertheless, there is no clear evidence that such approaches consistently provide superior geometrical accuracy, especially when applied to a limited number of patients. Therefore, a fixed HU threshold combined with manual refinement was considered an appropriate and controlled solution.

Previous studies have shown that variations in the Hounsfield Unit (HU) threshold can significantly affect the resulting geometry [35], but they do not provide a clear guideline for practical application.

The rationale behind the choice of the HU threshold is based on the difference in radiodensity between air and soft tissues: air exhibits a radiodensity of -1000 HU, while soft tissues generally fall within the range of -200 to -300 HU [15]. In principle, this contrast should allow for the automatic selection of the region of interest through a simple threshold-based operation. However, in practice, it is not possible to define a single threshold value that is universally valid.

Inter-patient anatomical variability, differences in mucosal thickness, and the presence of mucus can alter local radiodensity values, leading to reconstruction inaccuracies. As a result, manual supervision and adjustment of the HU threshold are required for each patient. This task should ideally be performed by someone with sufficient familiarity with nasal anatomy, who can assess whether the selected threshold produces a physiologically realistic geometry.

Two main types of errors may arise from an incorrect choice of the HU threshold. If the threshold is set too low (in absolute value), portions of the soft tissue may be incorrectly classified as air, leading to an artificial increase in the airway cross-sectional area and the formation of non-existent connecting passages between anatomically separate regions, as shown in Figure 2.1a (both upper nasal cavities are connected to the upper meatuses). Conversely, if the threshold is set too high, narrow airways may be erroneously filled with tissue, effectively occluding passages that are present in reality. This effect is particularly evident in the inferior meatus and in regions affected by mucus accumulation, as illustrated

in Figure 2.1c, see how in the lower left¹ meatus there is a occlusion, as if there was a turbinate hypertrophy.

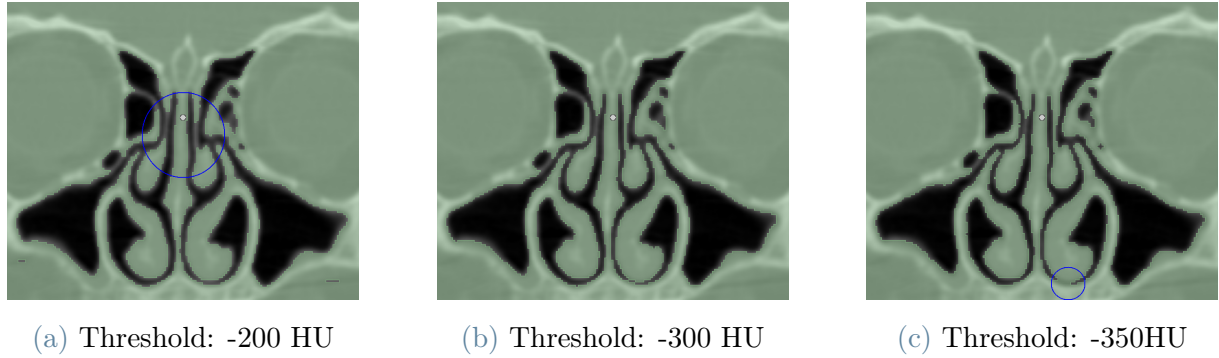


Figure 2.1: Effect of different HU thresholds applied to the same geometry. Patient 130, coronal section, CT resolution of 0.5 mm.

The final choice of a threshold value of -300 HU should therefore be regarded as a practical guideline derived from empirical experience rather than as a universally optimal value. In most cases, a threshold of -300 HU is sufficient to include all soft tissue structures, although it may occasionally result in the partial occlusion of narrow regions that require manual correction using the *Paint* tool.

From an operational standpoint, it is generally easier to locally widen an airway by manually (virtually) removing small amounts of tissue rather than to reconstruct missing anatomical features where tissue has been erroneously excluded. For this reason, a slightly more conservative threshold value (i.e., with a larger absolute value) is preferred, followed by localized manual refinement of the airway geometry.

2.2.2. Paranasal sinuses removal

In order to obtain a geometrically simpler and computationally more efficient mesh, all paranasal sinuses (frontal, maxillary, and sphenoidal) as well as the ethmoidal air cells are nearly always removed from the reconstructed geometry. This operation effectively reduces inter-patient anatomical variability, leading to improved performance in functional-mapping-based disease transfer between different subjects.

In addition, the removal of these volumes results in a reduction of approximately 40% in the total number of mesh points and about 20% in the number of internal points, as discussed in Chapter 3, section 3.1.3. Finally, the use of a simplified nasal geometry

¹Left and right are always considered by the patient prospective.

enables the application of simplified flow models based on the Hele-Shaw approximation, as investigated in a previous master's thesis [11].

In practice, the removal of the paranasal sinuses and ethmoidal air cells is conceptually straightforward but can be time-consuming. The most effective and efficient approach consists of using the *Erase* tool in 3D Slicer to isolate individual sinuses or air cells, followed by their removal through the *Keep largest island* operation.

The main challenge lies in determining the appropriate cutting boundaries. In some cases, the geometry is relatively simple. For example, maxillary sinuses can be removed by closing the ostium. In other cases, the process is significantly more complex. The ethmoidal labyrinth is composed of numerous interconnected air cells that lack a single, clearly defined channel connecting them to the main nasal cavity, making segmentation particularly challenging.

As a general guideline, it is important to preserve as much anatomically relevant geometry as possible while removing only the cavity volumes. Figure 2.2 illustrates the geometry before and after sinus removal in 3D Slicer, while Figure 2.3a shows the resulting geometry visualized in three dimensions.

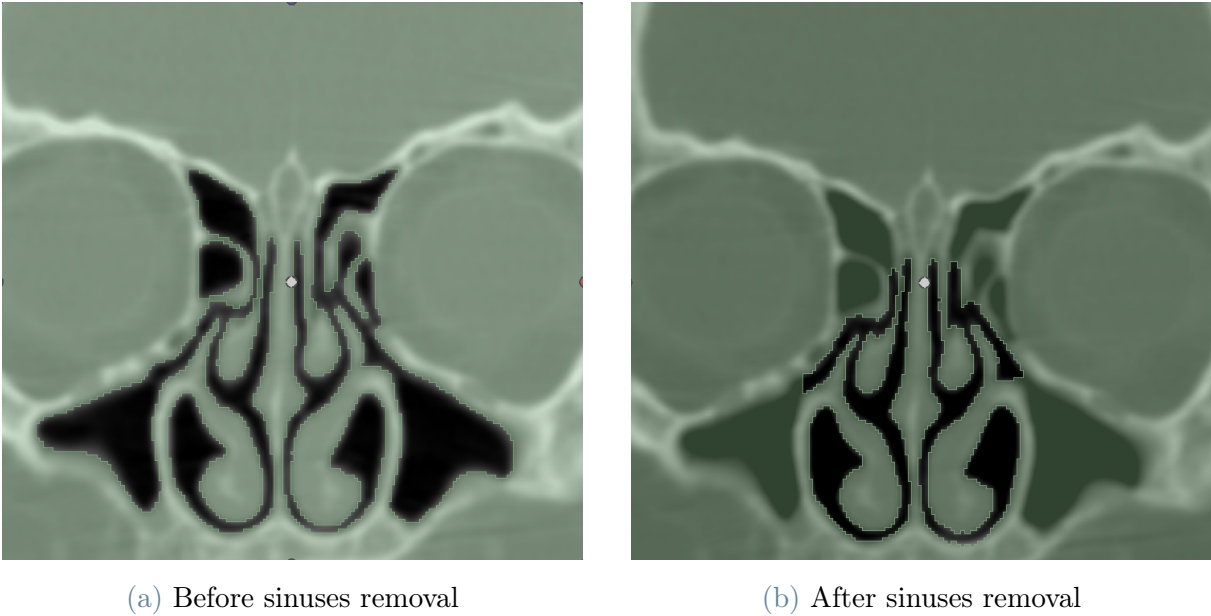


Figure 2.2: Comparison of the nasal geometry before and after sinus removal. The maxillary sinuses, the posterior portion of the frontal sinuses, and the most anterior ethmoidal air cells are visible in the original reconstruction. Patient 130, coronal section, CT resolution of 0.5 mm, HU = -300.

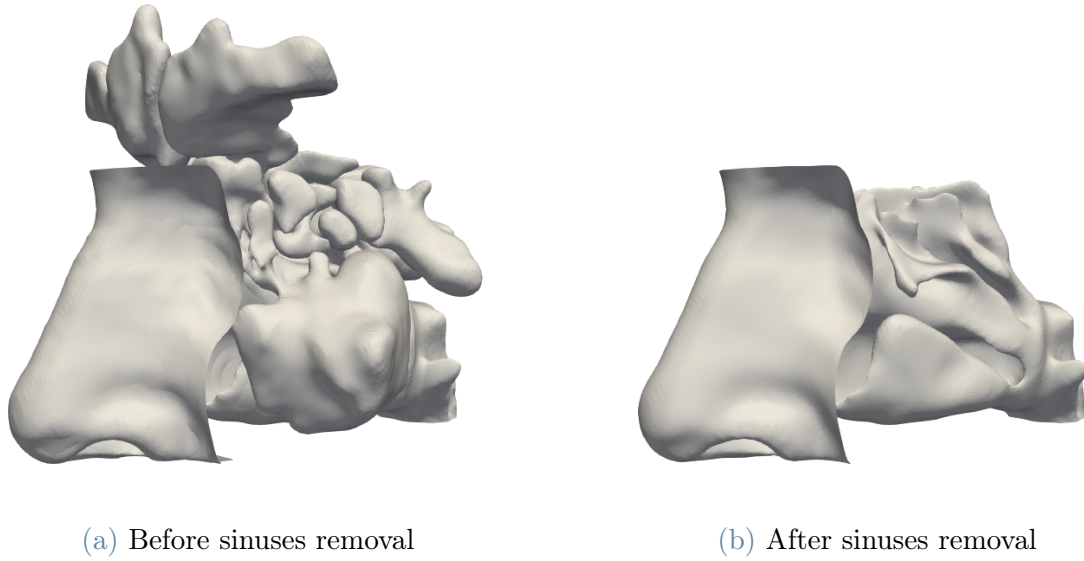


Figure 2.3: Comparison of the nasal geometry before and after sinuses removal. Patient 130, 3D view.

2.2.3. Additional settings

Before proceeding to the second stage of the reconstruction, it is necessary to briefly discuss additional parameters and methodological choices available during the reconstruction procedure, and to analyze their impact on the final mesh. Some of these choices have a minor influence on the reconstructed geometry, while others are expected to have a non-negligible effect that cannot be quantitatively assessed at this stage and require further investigation in future work.

Smoothing tool Within the *Segment Editor* environment, several smoothing options are available, all of which aim to reduce sharp edges and spikes in the segmented geometry. Although such tools may appear useful, their use proved to be harmful in the context of nasal airway reconstruction.

The smallest available smoothing kernel operates on a $3 \times 3 \times 3$ voxel cube. However, many anatomical features of the nasal cavity have characteristic dimensions smaller than three voxels. As a consequence, applying the smoothing tool frequently alters the real geometry by introducing non-existent occlusions or artificial connecting channels. In particular, smoothing almost systematically generates a connection between the two nasal cavities in their upper regions, where the septum is thinnest and the cavities are closest.

To reduce spike formation without compromising anatomical fidelity, the smoothing tool

is deliberately avoided. Instead, during manual editing, it is important to always keep at least four connected voxels, which is sufficient to prevent the generation of sharp spikes in the surface mesh. Any remaining geometric imperfections are subsequently addressed during the refinement stage in Blender.

Note: in the export menu there is another value, the *smoothing factor*. This number does not change substantially the reconstructed geometry (as the smoothing tool does) but smooths out the surface without adding or removing voxels. The default value is set to be 0.5. It is suggested to use 0.8 as it requires less work inside Blender.

Interpolation Since CT data are inherently discrete, each voxel is associated with a single grayscale intensity value. At small spatial scales, such as those characteristic of the nasal airways, this discretization can lead to a low-resolution or “pixelated” image. Interpolation is therefore required to fill the gaps between pixels, producing smoother transitions between grayscale values and, consequently, a smoother reconstructed surface mesh.

During the creation of the Region of Interest (ROI), different interpolation methods can be selected in 3D Slicer: *Linear* (default), *B-spline*, and *Windowed sinc*. According to the 3D Slicer documentation, the linear interpolator is classified as medium-quality, while the latter two are considered high-quality options.

From a purely visual inspection, no appreciable differences can be observed between linear and b-spline. However, a more detailed analysis reveals that the reconstructed geometries differ slightly, with discrepancies on the order of 10^{-2} mm. While such differences are small, they could have an impact on the nasal airflow, and a dedicated CFD analysis may be required to assess their effect on the simulated flow field.

No significant advantage was observed when using the *Windowed sinc* interpolator compared to *B-spline*. For this reason, it is suggested to use the *B-spline* interpolation method.

2.3. Blender

The second stage of the mesh creation process requires the use of Blender [48], a free and open-source 3D creation suite. This additional step is necessary because the STL file exported from 3D Slicer needs to be refined and the boundary condition surfaces have to be added.

In particular, three main steps are performed within Blender: the removal of the external walls automatically generated by 3D Slicer, localized smoothing and correction of the

mesh surface, and the construction of those walls that represent the external ambient. These walls are used to apply boundary conditions during the CFD simulations, see Sec. 3.1.2 for more details. Figure 2.4 illustrates the outcome of these three steps.

While the removal of the external walls is a relatively easy operation, manual surface smoothing requires greater ability with Blender tools. Localized refinement is performed using the *Smooth* brush inside the Sculpting environment to remove small surface imperfections. It is strongly recommended to perform the majority of the work in 3D Slicer, reserving Blender only for minor details corrections and refinement.

The final step consists of creating the inlet and outlet surfaces required for CFD simulations. These surfaces are generated as separate geometries with Blender and perfectly cut using the *Boolean modifier*. For all simulations presented in this work, boundary conditions are applied through a watertight, box-shaped volume positioned in front of the nose tip, representing the external ambient.

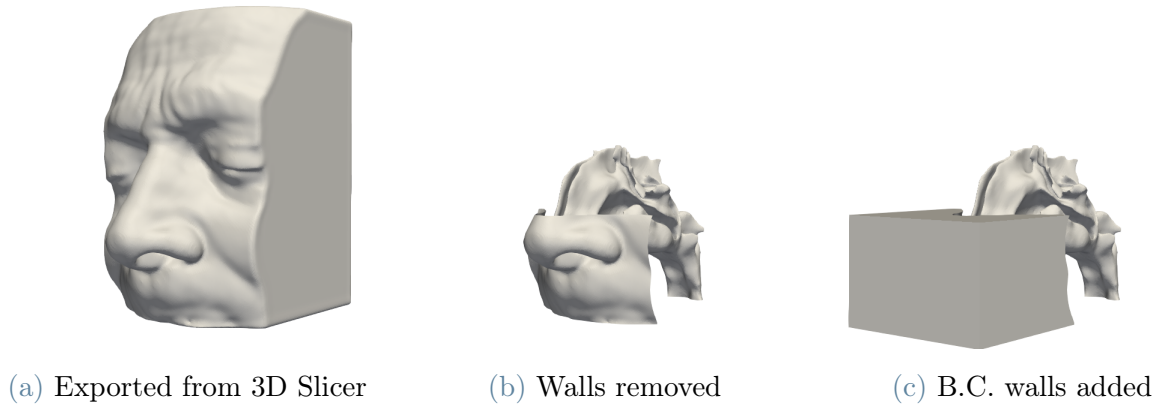


Figure 2.4: Three processing stages performed in Blender. Patient 81, geometry without sinuses, 3D view.

2.4. Use of the reconstruction procedure

The newly developed procedure was used to reconstruct a total of eight new nasal geometries from CT scan data. Five of these geometries correspond to almost completely healthy noses, with no significant pathological alterations, and can therefore be used to enrich the neural network database of the Open Nose project.

These models were generated starting from CT scans selected from a database of 320 elements made available to the Open Nose project by San Paolo Hospital. The scans in the database are indexed from 1 to 320. Patients numbered 41, 47, 69, 130, and 289 were

selected for this purpose.

Among them, patient 130 was selected as the reference baseline for all subsequently developed defects (see Appendix A) and for future functional-mapping defection projection, as it appears to be the healthiest and geometrically perfect.

Two additional geometries, derived from patients 116 and 145, were reconstructed but later discarded due to minor anatomical imperfections.

A final geometry, corresponding to patient 81, was reconstructed specifically for CFD simulation purposes, despite not being completely healthy. This geometry and its variants are analyzed in detail in Chapter 3, section 3.1.

3 | CFD Simulations

The aim of this chapter is to describe the choices adopted in the definition of the numerical simulations. In the first part, the three different geometries selected for the analysis are presented, and the main characteristics of the meshes are discussed. In the second part, the details of the simulation setup are described.

All simulations are performed using AMPHIBIOUS [1], the new DNS/LES solver developed by the Department of Aerospace Science and Technology of Politecnico of Milan. The computations are carried out on the Leonardo supercomputer at CINECA.

Three configurations were considered. For each configuration, both inspiration and expiration were simulated. All simulations were performed using both DNS and LES, for a total of twelve simulations, excluding those conducted for the convergence study.

3.1. Geometry

3.1.1. Nasal geometry

As already mentioned in the introduction, one of the the two objective of this work is to investigate whether, and to what extent, the flow field changes in a nasal geometry without paranasal sinuses compared to a complete anatomy. Any healthy subject would have been suitable for the analysis. However, an additional purpose of this study is to assess the influence of accessory maxillary ostia on the airflow both inside the maxillary sinus and in the connected meatus. For this reason, a patient presenting this anatomical feature was selected.

Accessory maxillary ostia (AMO) are a common anatomical variant located between the uncinate process and the inferior turbinate [23, 34]. This variation consists of an additional opening connecting the maxillary sinus to the main nasal cavity. Accessory maxillary ostia are common, occurring in up to 40% of the population. [9, 23]. It remains unclear whether the presence of an accessory ostium is only congenital or whether it can also be developed, for example as a consequence of chronic maxillary sinusitis leading to a

mucosal perforation [28].

For these reasons, a 78-year-old male subject presenting an accessory maxillary ostium was selected for the study. Henceforth, it is referred to as Patient 81. The patient exhibited a mild septal deviation and a frontal inferior nasal crest. The sinuses and turbinates were otherwise normal, with the exception of a slight posterior hypertrophy of the right inferior turbinate.

Two representative coronal CT scan slices of Patient 81 are shown in Figure 3.1. The image on the left (Fig. 3.1a) highlights the normal maxillary ostium, while the one on the right (Fig. 3.1b) shows the accessory maxillary ostium.

The nasal geometry was reconstructed from a CT scan with 384 DICOM images, with a sagittal and coronal resolution of 0.5 mm and an axial slice spacing of 0.6 mm. Segmentation was performed following the procedure developed and described in detail in Chapter 2.

Three different computational meshes were produced from the reconstructed geometry:

- a complete geometry without any modification,
- a geometry with the sinuses removed, and
- a geometry in which the accessory maxillary ostium was removed.

Effectively, the latter configuration is considered the reference “normal” geometry, while the geometry with the accessory ostium is treated as a variant. This strategy, removing an existing anatomical feature rather than manually adding one into a normal geometry, was considered more realistic, as the precise shape of an ostium is difficult to reproduce, whereas its removal can be performed in a straightforward way.

3.1.2. Boundary condition wall geometry

For all simulations presented in this work, boundary conditions are applied through a watertight, box-shaped volume positioned in front of the nose tip, representing the external ambient. The box three largest dimensions are 0.070052 mm $\times$ 0.06103 mm $\times$ 0.045965 mm. The approximate volume, obtained by multiplying the three given dimensions is $2.2 \times 10^{-4} \text{m}^3$. This value is overestimating the actual enclosed volume.

The nasal airway is closed downstream by an airtight surface located in the oropharynx, after the narrowing of the nasopharynx.

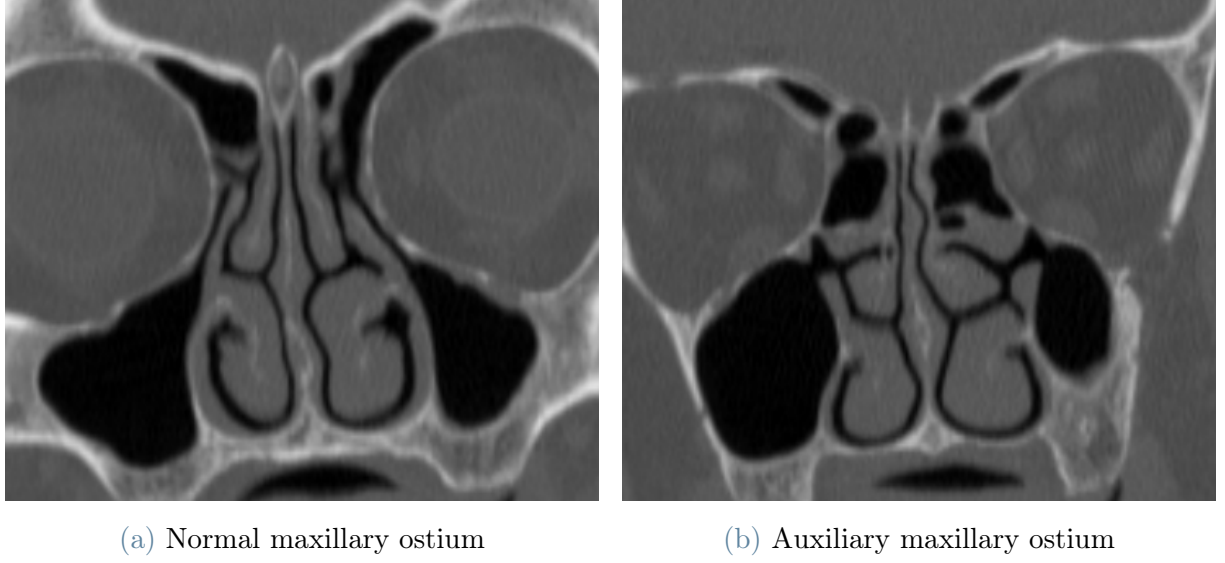


Figure 3.1: Two coronal section of Patient 81 CT scan showing the normal and the accessory ostia.

3.1.3. Final mesh characteristics

The three geometries were discretized using a uniform spatial resolution of 0.2 mm in all three directions. This resolution was selected to ensure adequate resolution of the flow field for the imposed pressure difference ΔP . Further justification of this choice and its impact on the solution is provided in Section 3.4.

Table 3.1 reports the main topological characteristics of the three 3D models. Information regarding the number of cells and surface points of the STL geometry was obtained using ParaView, while the total number of points and the number of internal flow points were extracted from the preprocessing log files generated by AMPHIBIOUS.

	Surface points	Cells	Total points	Internal points
Normal	311504	620049	139387488	28285961
No sinuses	202580	402181	88508700	22841685
Acc. Ost.	311255	619550	139387488	28282025

Table 3.1: Surface and domain characteristics for a resolution of 0.2mm

A more refined mesh with a uniform resolution of 0.1 mm was also generated. This mesh was used for convergence analysis and result verification. Its characteristics are nevertheless reported for completeness in Table 3.2. Since the surface mesh is unchanged, the number of surface points and cells is not repeated.

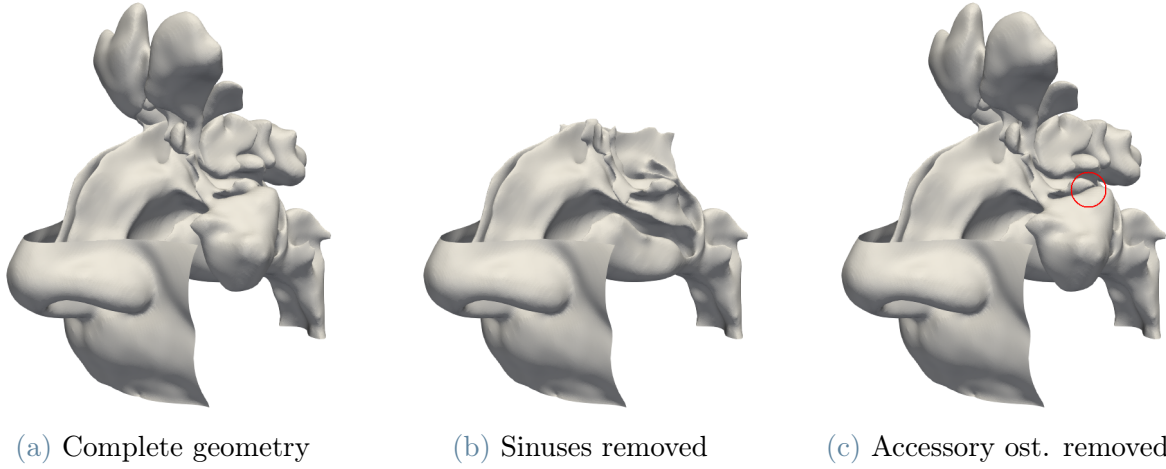


Figure 3.2: The three geometries generated from patient 81. From left to right: complete geometry with the double ostium, geometry without sinuses and geometry with the removal of the accessory ostium (physiologically the "normal" one). A zoomed in image showing the removal of the acc. ostum (red circle) can be found in Fig 3.3.

	Total n. of points	Internal points
Normal	1115099904	224634022
No sinuses	602229600	154690561
Acc. Ost.	1115099904	224665985

Table 3.2: Domain characteristics for a resolution of 0.1mm

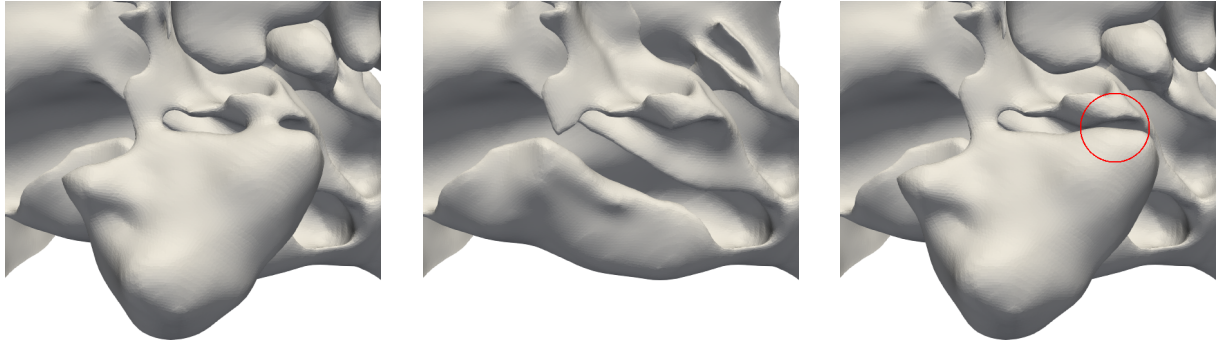
3.2. AMPHIBIOUS

Since AMPHIBIOUS is a relatively new software, and this work is the first of its kind to use it as the main solver for all simulations, it is important to introduce it to the reader and to presents its main features, as well as the reasons behind its selection.

AMPHIBIOUS is a finite-difference DNS solver for the incompressible Navier–Stokes equations, based on the immersed-boundary (IB) method. In more recent implementations, AMPHIBIOUS is also capable of performing LES simulations using two different models: WALE and Smagorinsky.

The source code, written in CPL [24], can be compiled for both CPU and GPU. Parallelization is achieved through MPI on CPUs and OpenACC on GPUs. AMPHIBIOUS is specifically designed to achieve high computational performance, reducing greatly computational time.

The software is split in three main modules: `preproc`, `solver` and `postproc`, and is



(a) Complete geometry

(b) Sinuses removed

(c) Accessory ost. removed

Figure 3.3: The three geometries generated from patient 81. Zoom on the left ostiomeatal region and focus on the maxillary ostia. From left to right: complete geometry with the double ostium, geometry without sinuses and geometry with only the normal ostium

controlled via a single input file `settings.in`.

All the additional details can be found on the official documentation on <https://www.amphibious.cfd/documentation/index.html> [1].

3.2.1. Theoretical background

The DNS solver implemented in AMPHIBIOUS is based on the immersed-boundary method (IBM) proposed by Luchini et al. [25] for the numerical solution of the incompressible Navier-Stokes equations.

Immersed-boundary methods rely on a Cartesian grid, in which the grid points do not generally coincide with the solid contours of the domain. This approach greatly simplifies grid generation, while significantly reducing the computational cost. These advantages are particularly relevant when simulating the flow inside or around geometrically complex geometries, such as the human nose. The effect of the body on the flow is effectively modelled as a correction in the form of an added forcing term in the N.S. equations.

In AMPHIBIOUS, the IB method is applied to the direct numerical simulation (DNS) of the incompressible Navier-Stokes equations written in primitive variables:

$$\begin{cases} \nabla \cdot \mathbf{u} = 0, \\ \frac{\partial \mathbf{u}}{\partial t} + (\mathbf{u} \cdot \nabla) \mathbf{u} + \nabla p - \frac{1}{Re} \nabla^2 \mathbf{u} = \mathbf{f}, \end{cases} \quad (3.1)$$

where $\mathbf{u}$ is the velocity vector, p is the pressure, Re is the Reynolds number, and $\mathbf{f}$ is a

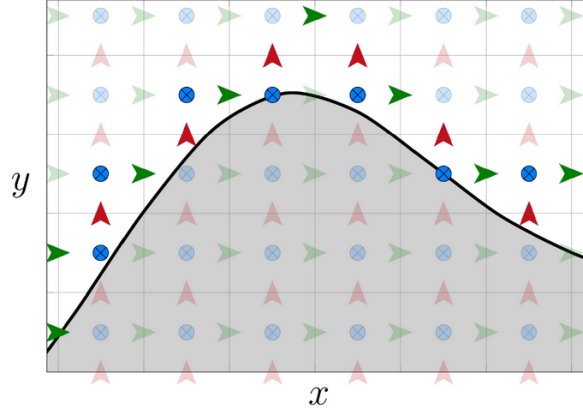


Figure 3.4: A solid body (gray background) with the staggered Cartesian grid. The collocation points for the velocity components in the x , y and z directions are drawn in green, red and blue. Dull colors indicate the fully internal/external points, and vivid colors indicate points where the IB correction is applied. Image from [25].

forcing term.

The equations are advanced in time using an incremental pressure-correction scheme coupled with an explicit temporal integration scheme. The momentum equation is first advanced without enforcing the incompressibility constraint. The velocity field is then projected onto a null-divergence space by solving a Poisson equation for the pressure increment, which is used to update both pressure and velocity. The Poisson equation is solved on the underlying Cartesian grid and is not modified by the presence of the immersed boundary [25].

Temporal discretization is performed using a three-stage Runge-Kutta scheme. The method achieves second-order accuracy in space regardless of the geometric complexity of the immersed boundary, while preserving the temporal order of accuracy.

Spatial discretization is carried out on a staggered Cartesian grid, where pressure is defined in the center of the cell and velocity components are defined at the corresponding cell interfaces, as shown in Fig. 3.4.

The principal novelty of this method is the implicit treatment, both in space and in time, of the immersed-boundary correction. This correction is concentrated exclusively on the central point of the Laplacian stencil, a property that allows an easier formulation allowing a natural extension to multiple spatial dimensions.

This feature leads to substantial simplifications in the code and results in a solver that is simple yet powerful, thanks to the implicit immersed-boundary correction.

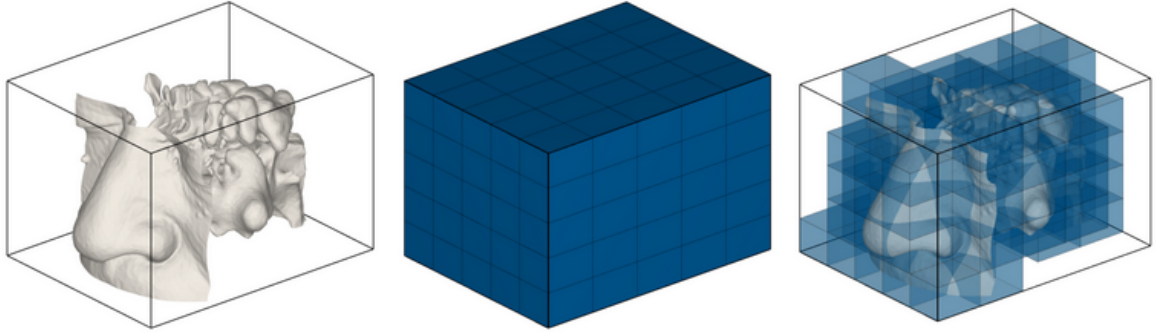


Figure 3.5: Decomposition of the bounding box into cubic subdomains and the pruning of empty cubes. The domain (left) is divided into $5 \times 5 \times 5$ cubes (center). Cubes containing no fluid points are removed (right), reducing the total number of subdomains from 125 to 91 and improving memory usage. Image from [1].

3.3. Simulation setup

3.3.1. General information and simplifications

For each of the three geometries, two main simulations were performed: one corresponding to **inspiration** and one to **expiration**. All simulations are statistically steady. For both cases, a pressure gradient of $|\Delta P| = 10 \text{ Pa}$ was imposed, adjusting the sign according to the desired flow direction. This value should generate a volumetric nasal flow that roughly corresponds to a respiration during light physical activity, such as walking [47]

Each simulation lasted a total physical time of 10.5 s . Time-averaged statistics were collected over the last 10 s , while the first 0.5 s were discarded in order to exclude the initial transient. An in-depth convergence analysis showed that the transient ends approximately after 0.2 s , making 0.5 s a conservative choice.

The kinematic viscosity of air was set to $\nu = 1.5 \cdot 10^{-5} \text{ m}^2/\text{s}$.

Simplifications were introduced. In particular, the nasal mucosa was not modeled and thermal fluxes were neglected. Temperature gradients were not considered in the analysis, as its implementation in AMPHIBIOUS is currently being developed.

Solver setup

An adaptive CFL strategy was adopted, with a maximum allowed value of $CFL_{\max} = 1$. The CFL number was updated every two iterations. A maximum time-step value of $dt_{\max} = 10^{-4} \text{ s}$ was imposed.

The pressure correction term was computed by iteratively solving the Poisson equation, with a total of 5 iterations at each time step.

LES

All Large Eddy Simulations were performed using the WALE (Wall-Adapting Local Eddy-viscosity) model. This choice is based on the comparative study presented in [43], where different LES turbulence models were evaluated for nasal cavity flows. The model constant was set to $C_m = 0.325$.

3.3.2. Boundary Conditions

Boundary conditions were imposed on both the external box and the oropharyngeal surfaces (see Sec. 3.1.2) by assigning the total pressure p_0 . For both inspiration and expiration, the total pressure at the external box was set to $p_0 = 0$ Pa. At the oropharynx, the total pressure was set to $p_0 = -10$ Pa during inspiration and to $p_0 = 10$ Pa for expiration, ensuring the correct flow direction.

AMPHIBIOUS allows to specify explicitly if boundary surfaces are either an inlet or an outlet. At inlet boundaries, the velocity and pressure components are imposed as follows. The velocity component normal to the boundary, u_n , and the two components parallel to the boundary, $u_{1,p}$ and $u_{2,p}$, satisfy:

- $\frac{du_n}{dn} = 0$
- $u_{1,p} = u_{2,p} = 0$
- $p = p_0 - \frac{1}{2}V^2$

At the outlet boundary, pressure and velocity conditions depends on the local presence of a backflow. Where there is no backflow, the entire velocity is imposed to be zero gradient, and the pressure is set to $p = p_0$. On the other hand, in the presence of backflow, the pressure condition is imposed as

$$p = p_0 - \frac{1}{2}V^2.$$

For inspiration the box in front of the nose is set to be the inlet and the oropharynx surface the outlet. The opposite is done for expiration.

3.3.3. Computational setup and costs

All simulations presented in this work were carried out via remote access to the *Leonardo* High Performance Computing supercomputer, hosted by CINECA and located at the Bologna DAMA Technopole. Leonardo is the pre-exascale Tier-0 supercomputer of the EuroHPC Joint Undertaking (JU).

In particular, since AMPHIBIOUS supports parallel computations on GPUs, the *Booster partition* of Leonardo was used. This partition is equipped with X2135 GPU blades based on NVIDIA Ampere A100-64 accelerators. The main hardware specifications are reported in Table 3.3. All details are publicly available on the official CINECA website [45].

Type	Specific
Models	Atos BullSequana X2135, Da Vinci single-node GPU
Racks	116
Nodes	3456
Processors/node	1x Intel Ice Lake Intel Xeon Platinum 8358
CPU/node	32
Accelerators/node	4x NVIDIA Ampere100 custom, 64 GiB HBM2e NVLink 3.0 (200 GB/s)
Local Storage/node (tmfs)	(none)
RAM/node	512 GiB DDR4 3200 MHz
Rmax	241.2 PFlop/s (TOP500)
Internal Network	200 Gbps NVIDIA Mellanox HDR InfiniBand – Dragonfly+ Topology
Storage (raw capacity)	106 PiB based on DDN ES7990X and Hard Drive Disks (Capacity Tier); 5.7 PiB based on DDN ES400NVX2 and Solid State Drives (Fast Tier)

Table 3.3: Leonardo hardware details

For the mesh with a resolution of 0.2 mm, the computational domain was divided into two subdomains ($nCubesX=2$, $nCubesY=1$, $nCubesZ=1$), each assigned to a single GPU. Consequently, each simulation required one compute node and only two out of the four GPUs each node has.

For the mesh with a resolution of 0.1 mm, the domain was divided into twelve subdomains ($nCubesX=3$, $nCubesY=2$, $nCubesZ=2$). In this case, three compute nodes were employed, each using four GPUs, for a total of twelve GPUs per simulation. See Sec. 3.3.4 for memory requirements of the two resolutions.

Tables 3.4 and 3.5 summarize the solver time and the corresponding computational cost expressed in effective CPU hours (CPUh), calculated according to the formula described

on the official CINECA website [45].

The billed hours B_H are calculated as

$$B_H = T \cdot N \cdot R \cdot C$$

where:

- T = elapsed time (in hours).
- N = number of nodes allocated.
- R = reserved resources per node (explained below).
- C = number of CPUs per node (depends on node architecture).

The R factor measures the fraction of node resources reserved by a job. It is defined as the maximum among all reserved resource types (e.g. the number of CPUs, GPUs), normalized by the total capacity of each respective resource on a single node.

$$R = \max \left\{ \frac{\text{Allocated CPUs}}{\text{CPUs in a node}}, \frac{\text{Allocated GPUs}}{\text{GPUs in a node}} \right\}$$

In this case each node has 32 CPUs and 4 GPUs (see Tab. 3.3).

INS.	Model	Solver t. (h)	Bill (CPUh)
Normal	DNS	3.92	63
	LES	4.30	69
No sin.	DNS	3.17	51
	LES	3.47	56
Acc. Ost.	DNS	3.97	63
	LES	4.37	69

Table 3.4: Simulation time and computational cost for inspiration cases at a spatial resolution of 0.2 mm, using 2 GPUs and 10.5 s of physical time. DNS simulations are highlighted in light blue. Billed hours are rounded to the nearest integer.

The high-resolution mesh ($\Delta x = 0.1$ mm), containing approximately eight times more points than the 0.2 mm mesh, required the use of twelve GPUs (3 full nodes) and resulted in a higher computational cost. For this reason, it was employed only for a convergence check, and simulations were performed only on the complete anatomy (with accessory ostium) over a physical time of 0.5 s, sufficient to exlude the transient.

It is possible to observe that, in general, expiration simulations are slightly more expensive than inspiration ones. On average, expiration cases require about 17% more computa-

EXP.	Model	Solver t. (h)	Bill (CPUh)
Normal	DNS	4.53	73
	LES	4.97	80
No sin.	DNS	3.82	62
	LES	4.15	67
Acc. Ost.	DNS	4.55	73
	LES	5.00	80

Table 3.5: Simulation time and computational cost for expiration cases at a spatial resolution of 0.2 mm, using 2 GPUs and a physical time of 10.5 s. DNS simulations are highlighted in light blue. Billed hours are rounded to the nearest integer.

tional time. Since the billing scales linearly with time time, they are also approximately 17% more expensive in CPUh.

Another interesting result, related to the implementation of the AMPHIBIOUS solver, is that LES simulations are only a little bit more expensive than DNS. On average, the computational cost of LES increases by approximately 9% with respect to DNS simulations.

3.3.4. Memory requirements

Memory usage and storage requirements for simulations with spatial resolutions of $\Delta x = 0.1$ mm and $\Delta x = 0.2$ mm are reported in Tables 3.6a and 3.6b.

Since the higher-resolution mesh was not integrated for the full simulation time, its statistical data are not available and are therefore not reported.

Although the 0.1 mm mesh contains exactly eight times the total number of grid points compared to the 0.2 mm mesh, the number of internal points is slightly lower (approximately 7.9 times larger, see Tabs. 3.1 and 3.2).

The memory required for storing the cube data structures (referred to as "Cubes" in Tab. 3.6b) is slightly more than eight times larger (a factor of 8.25). On the other hand, the memory occupation of the flow fields scales almost exactly with the number of internal points (factor ~ 7.9), confirming the expected proportionality.

A live monitoring of the GPU memory usage via `nvidia-smi` showed a requirement of approximately 17.1 GiB (≈ 18.4 GB) of GPU memory and about 17.5 GiB (≈ 18.8 GB) of RAM for the 0.2 mm simulation.

For the 0.1 mm mesh, the GPU memory consumption increased to approximately 138.0

GiB (≈ 148.2 GB), corresponding to a factor of 8.1 compared to the 0.2 mm case. The RAM usage was 135.3 GiB (≈ 145.3 GB), corresponding to a factor of 7.73.

Resolution [mm]	Cubes [GB]	Field [GB]		Stats [GB]	
		Binary	VTK	Binary	VTK
0.2	3.2	4.3	2.5	6.9	2.9
0.1	26.4	34	18	/	/

(a) Storage requirements for the two spatial resolutions.

Resolution [mm]	GPU [GiB]	RAM [GiB]
0.2	17.1	17.5
0.1	138.0	135.3

(b) GPU and RAM requirements for the two spatial resolutions.

Table 3.6: Memory, storage, GPU and RAM requirements for the two spatial resolutions.

3.4. Assessment of discretization

A study was performed to ensure that the initial transient was effectively excluded (convergence of the mean flow rate), that time-averaged quantities were sampled over a sufficiently long interval (time-averaged quantities convergence), and that the spatial resolution was fine enough to properly resolve small scale flow structures (grid convergence).

An additional comparison was carried out between the same simulations performed using DNS and LES, in order to quantify the differences between the two approaches.

3.4.1. Spatial resolution

To ensure that the spatial resolution is sufficient to resolve the small scale flow structures, the maximum ratio between the turbulent viscosity and the molecular viscosity, $\max(\nu_t/\nu)$, can be examined over the three-dimensional domain, assuming conventionally that a ratio equal or smaller than 1 means that all the structures are adequately resolved.

Tables 3.7 and 3.8 show that, for a spatial resolution of $\Delta x = 0.2$ mm and the imposed pressure gradient of 10 Pa, the ratio ν_t/ν remains below 1 in all inspiration simulations. This confirms that the chosen resolution is sufficient for these cases.

For expiration simulations, the maximum value of ν_t/ν is slightly bigger than 1 in some geometries. However, this is not a problem, since values of ν_t equal to or larger than ν

are observed only in few isolated cells, as expected. This behavior is shown in Fig. 3.6, where the red contour $\nu_t/\nu = 1$ is small and localized.

Note that the turbulent viscosity ν_t is computed a posteriori for the DNS simulations, while it is directly calculated during runtime in the LES case. As expected, the ratio ν_t/ν is generally lower for LES simulations.

INS.		$\max(\nu_t)$	$\max(\nu_t/\nu)$
Normal	DNS	9.81e-06	0.62
	LES	7.74e-06	0.52
No sin.	DNS	9.75e-06	0.65
	LES	9.49e-06	0.63
Acc. Ost.	DNS	9.49e-06	0.63
	LES	8.37e-06	0.62

Table 3.7: $\max(\nu_t)$ and $\max(\nu_t/\nu)$ during inspiration. Resolution of 0.2 mm

EXP.		$\max(\nu_t)$	$\max(\nu_t/\nu)$
Normal	DNS	1.79e-05	1.19
	LES	1.79e-05	1.19
No sin.	DNS	1.60e-05	1.07
	LES	1.26e-05	0.84
Acc. Ost.	DNS	2.25e-05	1.50
	LES	1.59e-05	1.06

Table 3.8: $\max(\nu_t)$ and $\max(\nu_t/\nu)$ during expiration. Resolution of 0.2 mm

An additional convergence check was performed using the high-resolution mesh with $\Delta x = 0.1$ mm. An expiration simulation was carried out on the complete geometry and the results were compared with those obtained using the lower-resolution mesh. The numerical comparison of the total flow rate is reported in Table 3.9.

Resolution [mm]	Q_{mean} [m^3/s]	σ [m^3/s]	σ %	Δ [m^3/s]	Δ %
0.2	$2.388 \cdot 10^{-4}$	$1.27 \cdot 10^{-7}$	0.05%	/	/
0.1	$2.374 \cdot 10^{-4}$	$4.77 \cdot 10^{-7}$	0.2%	$1.495 \cdot 10^{-6}$	0.6 %

Table 3.9: Estimated mean flow rate Q_{mean} for the expiration simulation in the complete geometry (with accessory ostium), standard deviation on the estimated mean and the difference. Comparison between the 0.2 mm and 0.1 mm mesh resolutions.

The relative difference in the mean volumetric flow rate between the two resolutions is below 1% and the absolute difference between the mean values ($1.5 \cdot 10^{-6} m^3/s$) is larger than the estimated standard deviation of the mean in both cases.

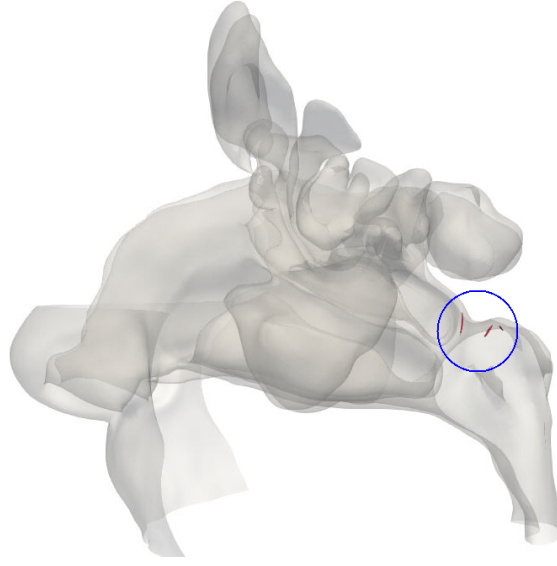


Figure 3.6: Ratio between turbulent and molecular viscosity for the full geometry (with accessory ostium). Red isosurface: $\nu_t/\nu = 1$.

Mean value estimate and its standard deviations were computed using the unbiased adaptive batch means method [38].

This indicates that the solution is only weakly sensitive to further mesh refinement and that the $\Delta x = 0.2$ mm resolution provides a sufficiently grid-independent estimate of the global flow quantities.

3.4.2. Averaging time

To ensure that a sufficiently long time was discarded before sampling quantities, the temporal evolution of the mass flow rate was analyzed. Figure 3.7 shows the evolution of the nasal flow during an inspiration simulation for the three geometries considered. It can be observed that the initial transient and the largest oscillations decay within approximately 0.2 s. Therefore, discarding the first 0.5 s of simulation time is a conservative and safe choice.

The convergence of time-averaged quantities was then assessed by comparing simulations of different duration. Three cases were considered: a 2.5 s simulation with quantities averaged over the last 2 s, a 4.5 s simulation with the same quantities averaged over the last 4 s and a 10.5 s simulation with 10 s of averaging time.

Figure 3.8 compares the magnitude of the time-averaged velocity U_{mag} for inspiration and expiration, respectively, between simulations with a physical duration of 4.5 s and

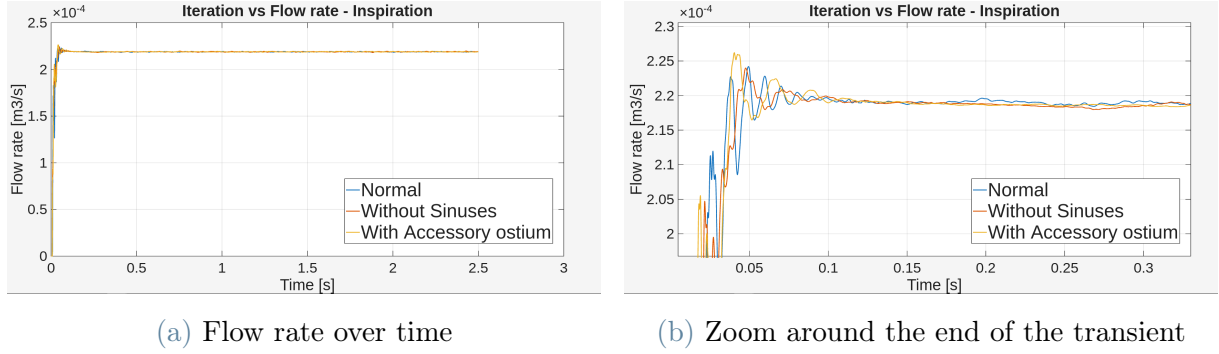


Figure 3.7: Temporal evolution of the nasal flow rate during an inspiration simulation of 2.5 s. The image on the right shows a zoomed view around the end of the transient.

10.5 s. Since the two averaged fields are visually indistinguishable, only the results of the 10.5 s simulations are reported, together with the absolute difference between the averages computed over 10.5 s, 4.5 s and 2.5 s.

The comparison is performed on a sagittal section of the left nasal cavity, shown in Fig. 4.1.

Since the differences the difference observed between the 10 s and 4 s simulations are very small for both inspiration and expiration cases (with the biggest variation in the volume of U_{mag} being around 2.61%) it is safe to say that the chosen averaging time of 10 s is sufficient to obtain statistically converged average quantities.

3.4.3. DNS vs LES

An additional analysis was performed comparing results obtained using DNS and LES. The comparison was carried out comparing the net volumetric nasal flow rates over the 10.5 s simulations, as well as on the time-averaged velocity magnitude U_{mag} .

Since the absolute fields obtained with the two approaches are almost identical, only the absolute differences are shown in Figures 3.9.

Table 3.10 reports the relative differences in the net volumetric flow rate computed using DNS and LES. The differences are below 0.1% for both inspiration and expiration, and can be considered negligible.

The comparison was performed on the magnitude of the time-averaged velocity field U_{mag} , averaged over the last 10 s of simulation time.

Figures 3.9 show the differences between DNS and LES results for inspiration and expiration. The two approaches generate very similar mean velocity fields (not shown here)

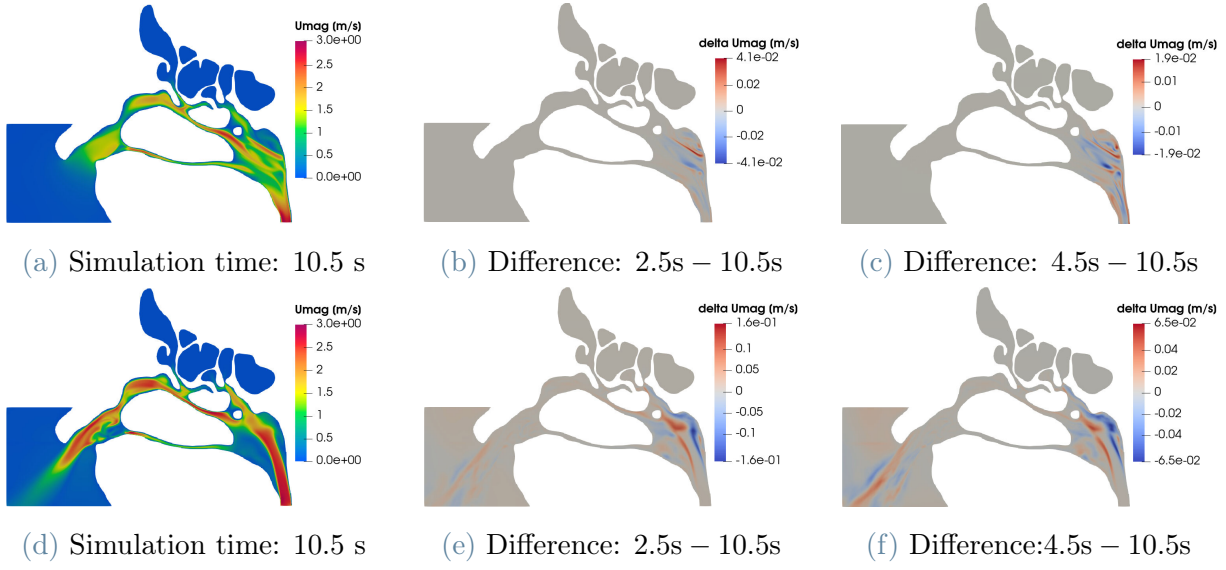


Figure 3.8: Sagittal section of simulations for the complete geometry (with accessory ostium). Time-averaged velocity magnitude U_{mag} [m/s] computed over 10.5 s (left), absolute difference between the averages obtained over 10.5 s and 2.5 s (center), absolute difference between the averages obtained over 10.5 s and 4.5 s (right). Top row: inspiration. Bottom row: expiration.

	DNS [m^3/s]	LES [m^3/s]	Δ %
Inspiration	2.188e-04	2.187e-04	0.05%
Expiration	2.388e-04	2.386e-04	0.08%

Table 3.10: Net volumetric flow rate for the complete geometry (with accessory ostium) during inspiration and expiration, comparing DNS and LES simulations.

with only minor local discrepancies up to 3% considering the maximum difference over the maximum U_{mag} in the DNS case. These differences are confined to the nasopharyngeal region and, in the expiration case, to the outward jet at the outlet and the volume in its proximity. These results are not surprising, since those are areas where the physics is richer and where k is higher, see Fig. 3.10.

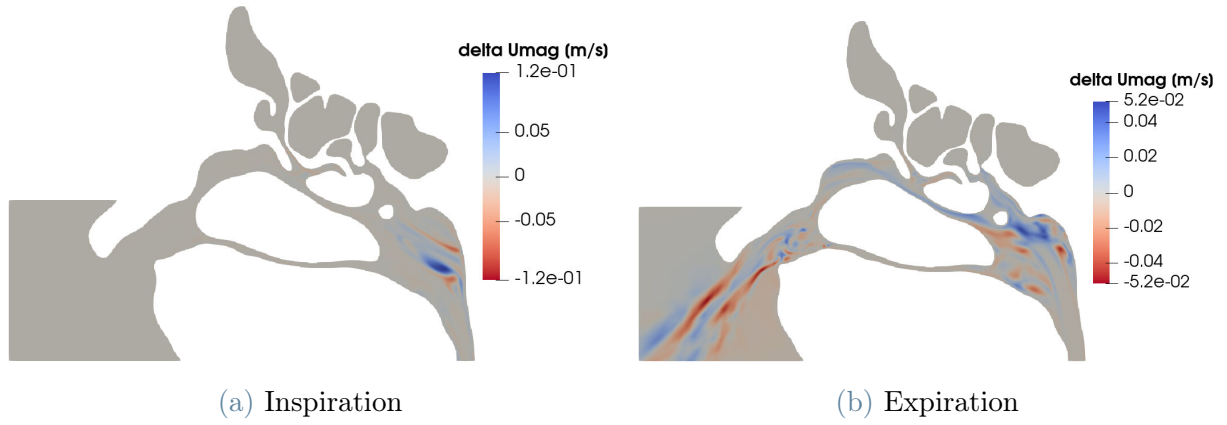


Figure 3.9: Sagittal section of inspiration and expiration simulations for the complete geometry (with accessory ostium). Difference of the time-averaged velocity magnitude U_{mag} for DNS and LES simulations. For the total DNS field see Fig. 3.8 a.

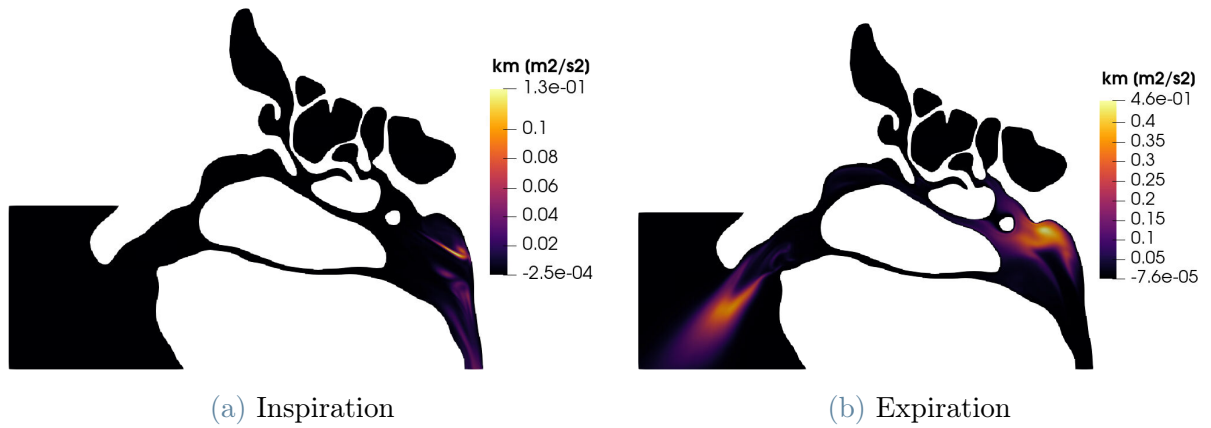


Figure 3.10: Sagittal section of inspiration and expiration simulations for the complete geometry (with accessory ostium). Turbulent kinetic energy k [m^2/s^2] for DNS simulations.

4 | Results

4.1. Overview

The results are now presented as follows. First, it is shown that the three nasal geometries are functionally the same from a global point of view. Subsequently, local differences are analyzed in detail.

Four main types of slices are used throughout the analysis: one sagittal slice taken in the middle of the left nasal cavity (Figure 4.1), and three coronal sections, referred to as S1, S2, and S3 (Figure 4.2) that highlight the normal ostium, the accessory ostium and ethmoidal cells, respectively.

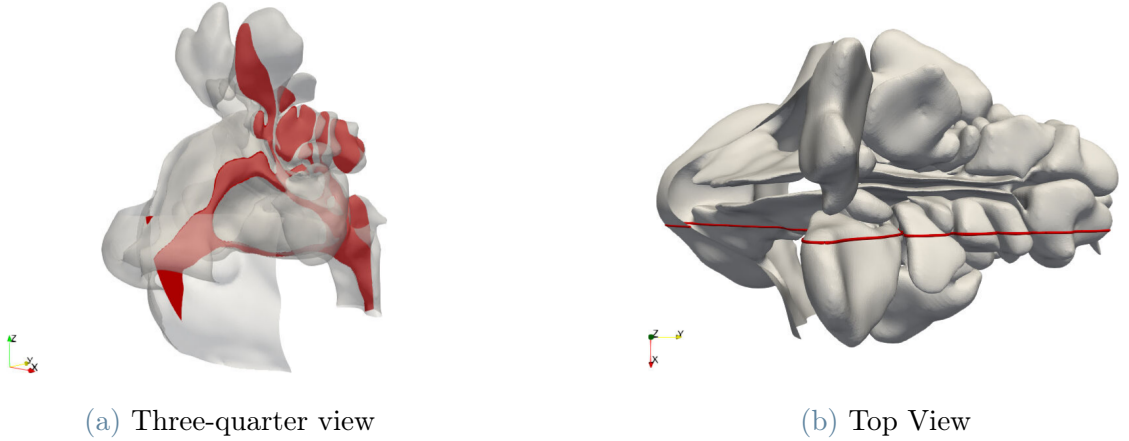


Figure 4.1: Sagittal section used for global visualization. The slice highlights the nasal valve, the lower and middle meati, the left frontal and sphenoidal sinuses, and some ethmoidal cells. The adopted reference system is shown.

4.2. Global analogy

The quantities used for the comparison are the total flow rate Q and the overall nasal resistance, defined as $R = \frac{\Delta P}{Q}$, where the pressure drop is calculated as $\Delta P = P_{\text{in}} - P_{\text{out}}$. Since the external pressure is fixed to zero in the simulations, only the internal pressure

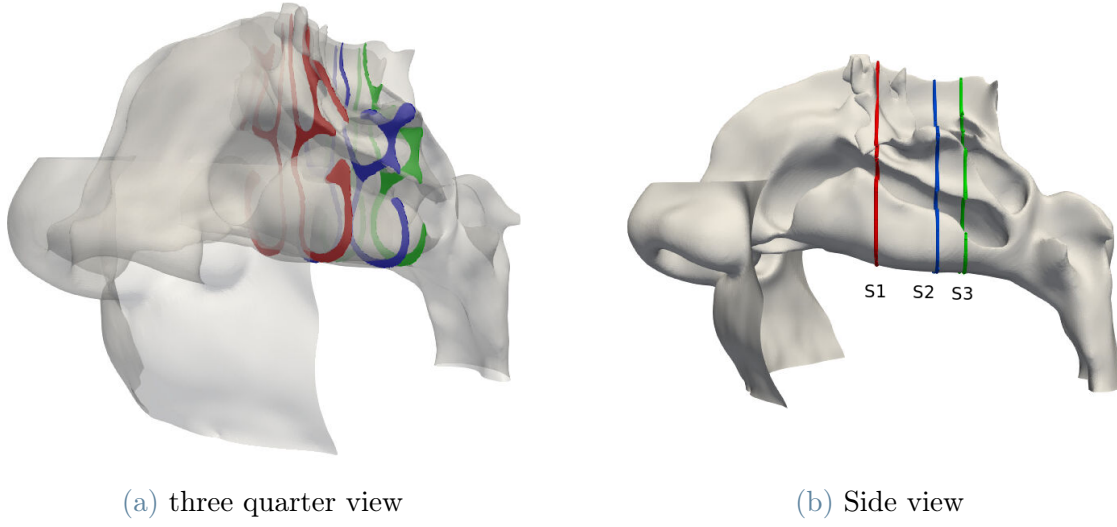


Figure 4.2: Coronal sections used for local analysis. Section S1 (red) cut through the normal ostium and frontal sinus, S2 (green) the accessory ostium, and S3 (blue) the ethmoidal cells. The geometry without sinuses is used in the figure to improve readability.

field is sampled. In particular, the pressure is virtually probed at the center of the nasopharynx, as commonly done in studies of this kind [31].

Table 4.1 reports the results obtained during inspiration simulations. The total flow rate is identical for the three geometries up to the second decimal digit. The value of approximately 219 mL/s is consistent with the pressure difference and it is coherent with experimental measurements as a flow rate during light physical activity [47].

The nasal resistance remains unvaried, with a value of $R = 0.028$ Pa/(mL/s), which lies within the normal range for healthy adult patients, see [14, 31].

Table 4.2 presents the corresponding quantities during expiration. Even though the total flow rate is slightly higher than in the inspiration case, it is essentially identical in the three configurations, as expected.

INS.	ΔP_m [Pa]	Q [mL/s]	R [Pa/(mL/s)]
Normal	−6.179	218.84	0.028
No sinuses	−6.175	218.82	0.028
Acc. ostium	−6.172	218.81	0.028

Table 4.1: Inspiration simulation results. Total flow rate Q (positive inward), mean pressure drop ΔP_m measured at the center of the nasopharynx, and nasal resistance R .

Time-averaged velocity magnitude U_{mag} is also used to support the observation that the global velocity distribution remains unchanged among the three cases. Figure 4.3

EXP.	ΔP_m [Pa]	Q [mL/s]	R [Pa/(mL/s)]
Normal	6.970	239.15	0.029
No sinuses	6.989	238.95	0.029
Acc. ostium	6.966	239.07	0.029

Table 4.2: Expiration simulation results. Total flow rate Q (positive outward), mean pressure drop ΔP_m measured at the center of the nasopharynx, and corresponding nasal resistance R .

(inspiration in the top row, expiration in the bottom) highlights the almost identical velocity fields, confirming what was already observed in terms of flow rate and nasal resistance.

Additional quantities (e.g. P_m , k_m , ...) can be seen in Appendix B.

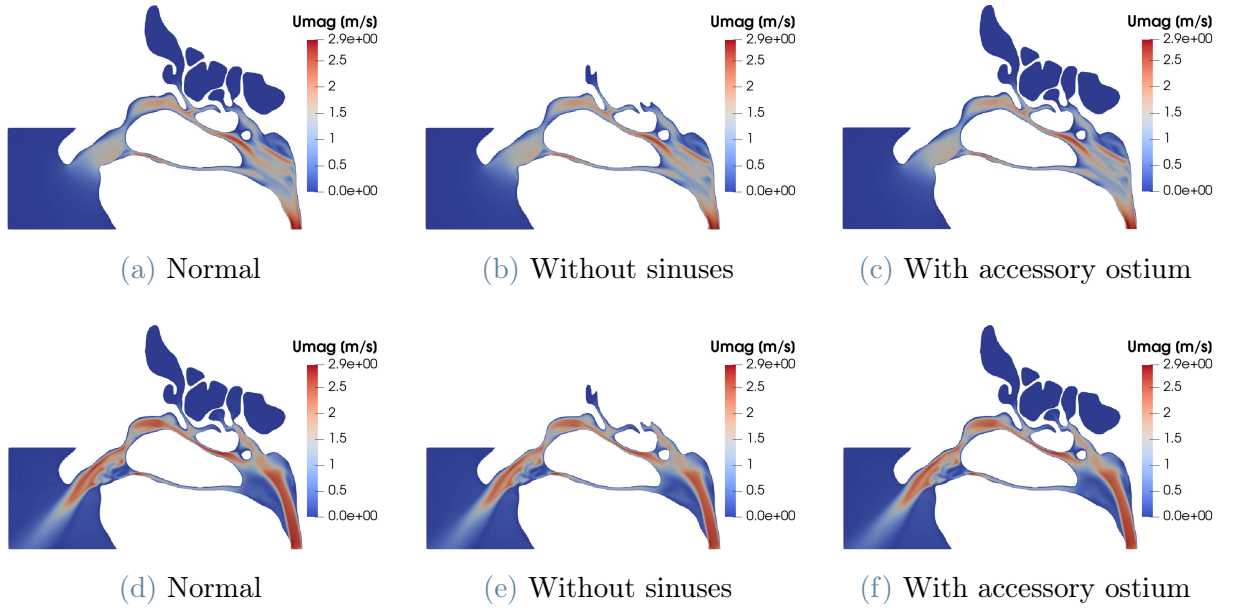


Figure 4.3: Time-averaged velocity magnitude U_{mag} [m/s] in sagittal section. Top row: inspiration. Bottom row: expiration. The three configurations show almost identical velocity distributions.

4.3. Local differences

Differences in the flow field are observable only at a local scale, in regions where the geometry was changed. The analysis is therefore divided into two parts: firstly, the sinuses removal from the normal geometry, and secondly, the addition of the accessory ostium to the same geometry

Coronal sections S1, S2, and S3 are used to highlight these effects on the local field.

4.3.1. Sinuses removal

The removal of the sinuses has a limited influence on pressure, even locally. For this reason, the time-averaged velocity magnitude U_{mag} is used to see the local differences. Additional quantities are reported in Appendix B.

Figure 4.4 shows the velocity magnitude in section S1 for the normal geometry, the geometry without sinuses, and their difference, for both inspiration and expiration. Local differences up to 0.6 m/s can be seen in the ostiomeatal complex, near the tip of the uncinate process. Smaller differences are found in the upper part of the section, where frontal sinuses were removed. On the other hand, the removal of the maxillary sinus has a almost no effect on the velocity field near the closed ostia.

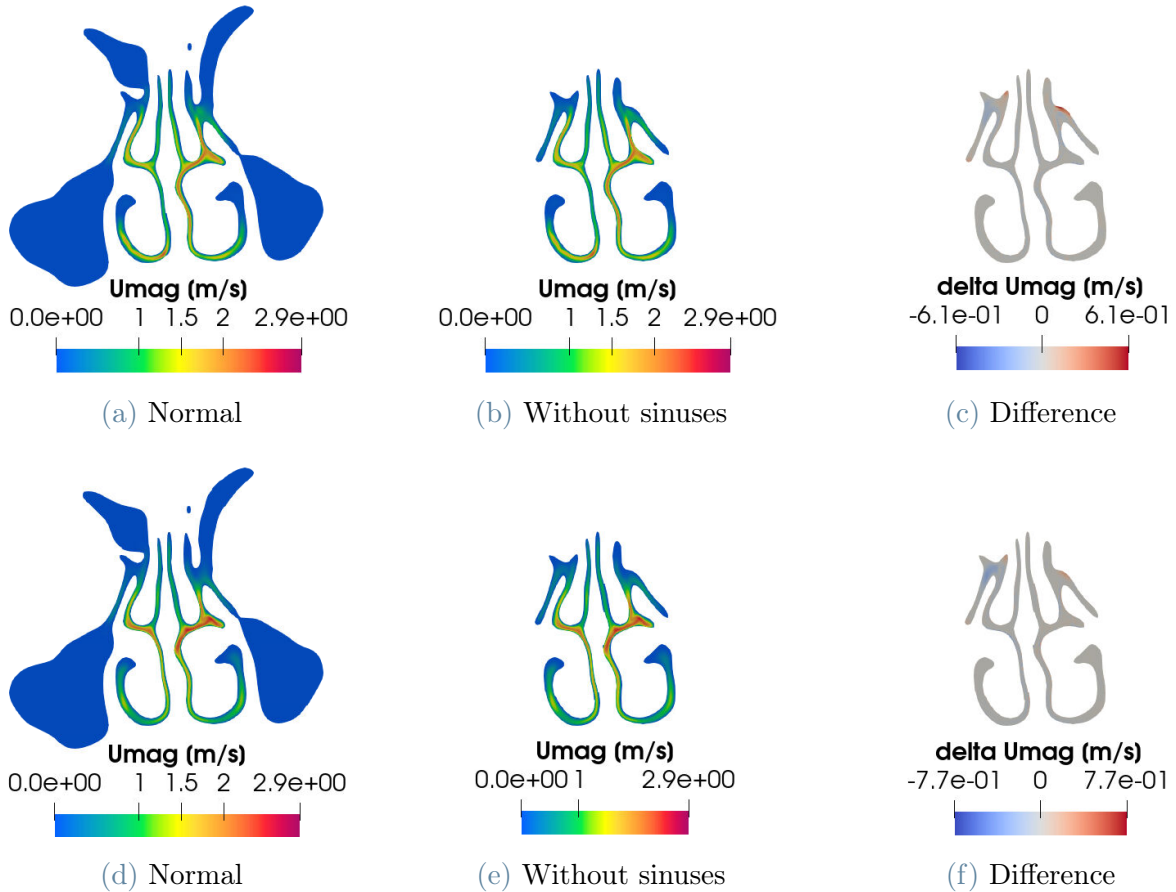


Figure 4.4: Section S1 showing the time-averaged velocity magnitude U_{mag} [m/s] for the normal geometry, the geometry without sinuses, and their difference. Extrema of the colorbar refer to the slice, not to the volume. Top row: inspiration. Bottom row: expiration.

Figure 4.9 shows the same comparison for section S3. Local differences of comparable magnitude are observed in regions where the ethmoidal cells were removed. These differences are more pronounced during expiration.

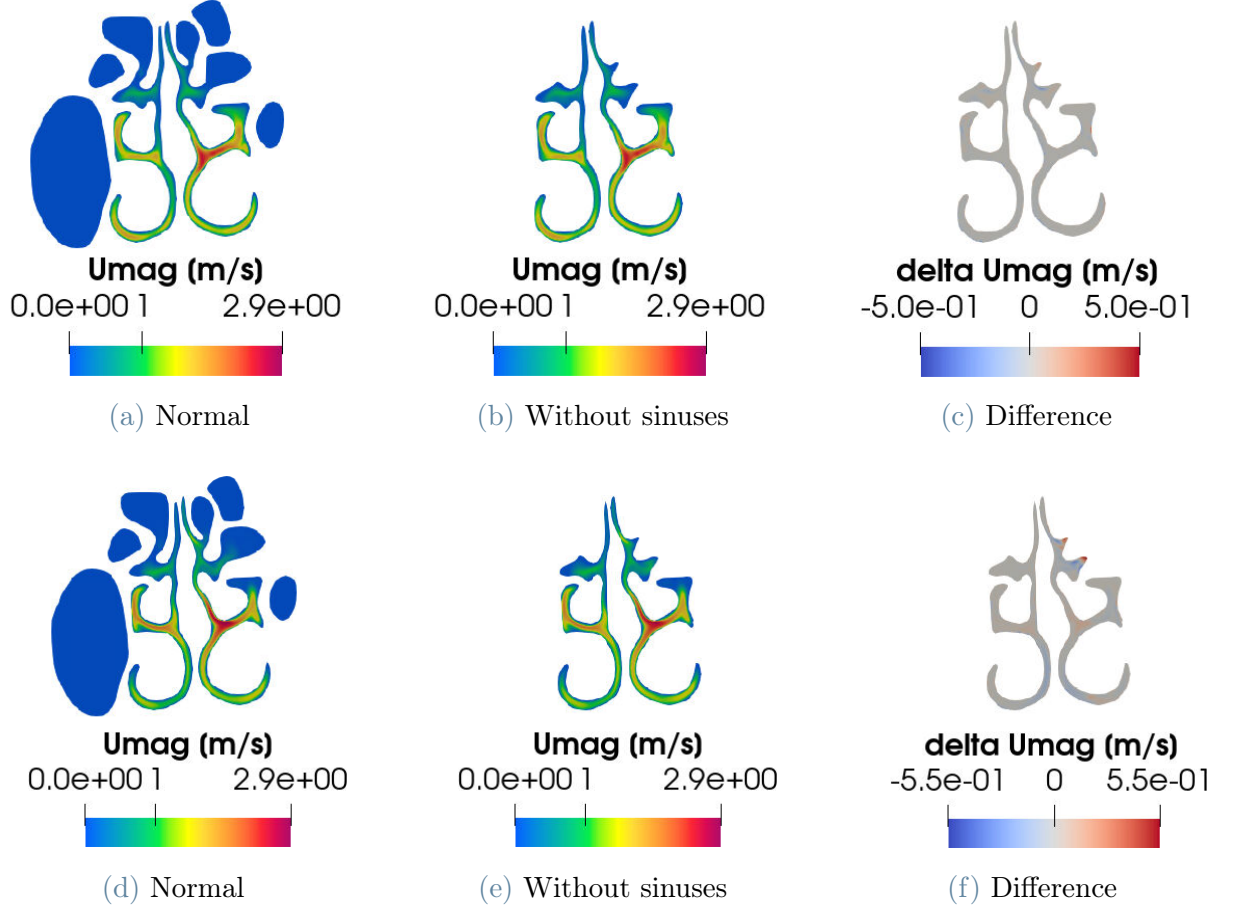


Figure 4.5: Section S3 showing the time-averaged velocity magnitude U_{mag} [m/s] for the normal geometry, the geometry without sinuses, and their difference. Top row: inspiration. Bottom row: expiration.

Additional quantities that help to understand the local influence of sinus removal are the root mean square (RMS) values of the velocity fluctuations, $\sqrt{u'^2}$, $\sqrt{v'^2}$, and $\sqrt{w'^2}$, as well as their combined contribution represented by $2k$. In the following discussion, only the quantities considered most representative are shown, while additional figures are reported in Appendix B.

Figure 4.6 shows $\sqrt{u'^2}$ for the normal geometry, the geometry without sinuses, and their difference in section S1 during inspiration. Image (c) highlights the effect of sinus removal: the largest differences (approximately 3.2×10^{-3} m/s) are observed in the proximity of the closed maxillary and frontal ostia. By comparing panels (a) and (b), it can be observed

that the geometry without sinuses present reduced fluctuation levels in these regions.

A similar behaviour is observed in Fig. 4.7, where $\sqrt{w'^2}$ is reported for expiration in section S3. Image (c) shows localized peaks in $\Delta\sqrt{w'^2}$, associated with the removal of the ethmoidal cells. Also in this case, a comparison between panels (a) and (b) shows that the sinus-less geometry presents lower oscillation intensity compared to the actual anatomical configuration.

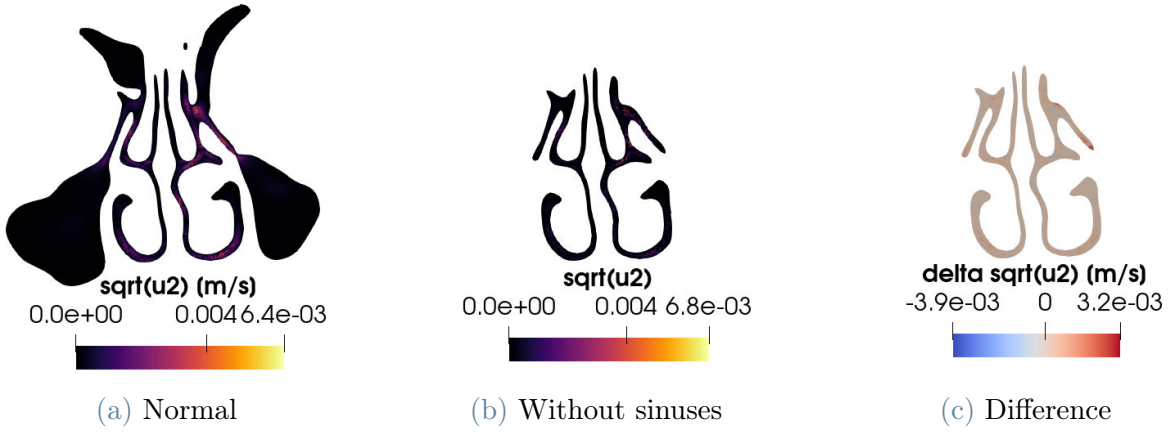


Figure 4.6: Root mean square of the velocity fluctuations in $\hat{i}$ direction, $\sqrt{u'^2}$. Section 1, with and without sinuses, inspiration simulation.

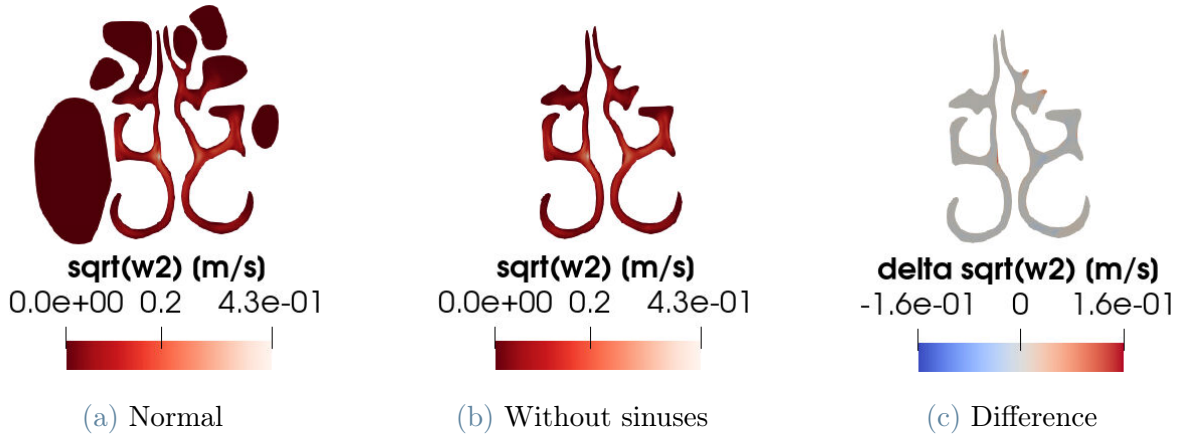


Figure 4.7: Root mean square of the velocity fluctuations in $\hat{k}$ direction, $\sqrt{w'^2}$. Section 3, with and without sinuses, expiration simulation.

Finally, an ulterior local difference emerges when looking at the local pressure drop ΔP across the ostia. By measuring the pressure immediately upstream and downstream of each ostium along a longitudinal direction, it is possible to assess how the local pressure drop changes with the sinuses removal. Table 4.3 reports the values obtained for the maxillary, frontal, and sphenoidal sinuses ostia. An interesting results is that the variation

in ΔP may change sign and magnitude depending on the specific geometry or the ostium considered. In most cases, the presence of the sinuses is associated with a larger ΔP across the ostium, while the removal of the sinuses leads to a smaller pressure drop.

However, this does not happen everywhere. For instance, the difference of pressure drop measured across the left maxillary sinus (first row of Table 4.3) shows the opposite trend, with a larger ΔP when the sinus is removed. An opposite behavior is observed for the right maxillary sinus (not reported in the table), where the difference in ΔP is smaller in but positive, with a value of approximately $0.01Pa$.

These results show that local pressure-drop measurements are highly sensitive to both the local geometry and the exact location of the probed points. Nevertheless, it is reasonable to conclude that the removal of the sinuses does change the local pressure drop across the ostia, even though it is difficult to predict beforehand the magnitude and the sign of this change.

	ΔP Normal	ΔP No sin.	Abs. difference	Δ %
Max. sinus	0.613	0.659	-0.05	- 7.5 %
Front. sinus	0.495	0.473	0.022	4.4 %
Sphen. sinus	0.336	0.329	0.007	2.1 %

Table 4.3: Local pressure drop ΔP (in Pa) measured across the ostia in a longitudinal direction for the maxillary, frontal, and sphenoidal sinuses. Values are reported for the normal geometry and for the geometry without sinuses, together with their difference and the percentage of variation. Measurements refer to the left-side sinuses only.

4.3.2. Accessory ostium

As expected, the presence of a second ostium introduces an additional path for the airflow, allowing air to enter through one ostium and exit through the other. During inspiration, the flow enters the sinus through the normal ostium and exits through the accessory ostium. The opposite behaviour is observed during expiration.

The mean volumetric flow rate through the ostia was computed in ParaView by integrating the normal velocity over local section (see Fig. 4.8). The measured areas of the four sections are reported in Table 4.4. Tables 4.5a and 4.5b report the volumetric flow rates during inspiration and expiration, respectively.

For comparison, Table 4.6 reports the volumetric flow rate through the normal ostium in the physiological normal geometry, where only one ostium is present. As expected, the net flow through the ostium is almost zero. The small error can be attributed to

numerical and geometrical uncertainties, such as segmentation inaccuracies from the CT scan, discretization effects, and the numerical integration procedure used in ParaView. Another source of error could be the number of iterations done inside AMPHIBIOUS to solve the pressure equation. A higher number of iterations may lower this error.

	Norm. ost. [m^2]	Acc. ost. [m^2]
Left	6.05e-06	1.18e-05
Right	1.76e-05	8.88e-06

Table 4.4: Area value of the four maxillary ostia.

Ins.	Norm. ost. [ml/s]	Acc. ost. [ml/s]	Δ [m^3/s]
Left	1.48	-1.49	8.18e-09
Right	2.16	-2.15	1.48e-08

(a) Mean volumetric flow through the ostia during inspiration. Section normals are oriented toward the maxillary sinuses.

Exp.	Norm. ost. [ml/s]	Acc. ost. [ml/s]	Δ [m^3/s]
Left	-1.47	1.49	2.58e-08
Right	-2.10	2.06	3.36e-09

(b) Mean volumetric flow through the ostia during expiration. Section normals are oriented toward the maxillary sinuses.

Table 4.5: Mean volumetric flow through the maxillary ostia during inspiration and expiration for normal and accessory ostium configurations.

	Inspiration [m^3/s]	Expiration [m^3/s]
Left	6.287e-11	1.009e-09
Right	3.527e-10	6.438e-10

Table 4.6: Mean volumetric flow through the normal ostia during inspiration and expiration (absolute value). Physiology with just the normal ostium.

Figure 4.9 shows the time-averaged velocity magnitude in section S2 for the normal geometry, the geometry with an accessory ostium, and their difference, for both inspiration and expiration. Local differences of up to 0.5 m/s are observed in the regions where the ostia were added. This is consistent with the non-zero volumetric flow previously calculated.

Also in this case, the root mean square (RMS) of the velocity fluctuations is analysed. Figures 4.10 and 4.11 highlight the differences in $\sqrt{u'^2}$ during inspiration (Fig. 4.10) and in $\sqrt{w'^2}$ during expiration (Fig. 4.11).

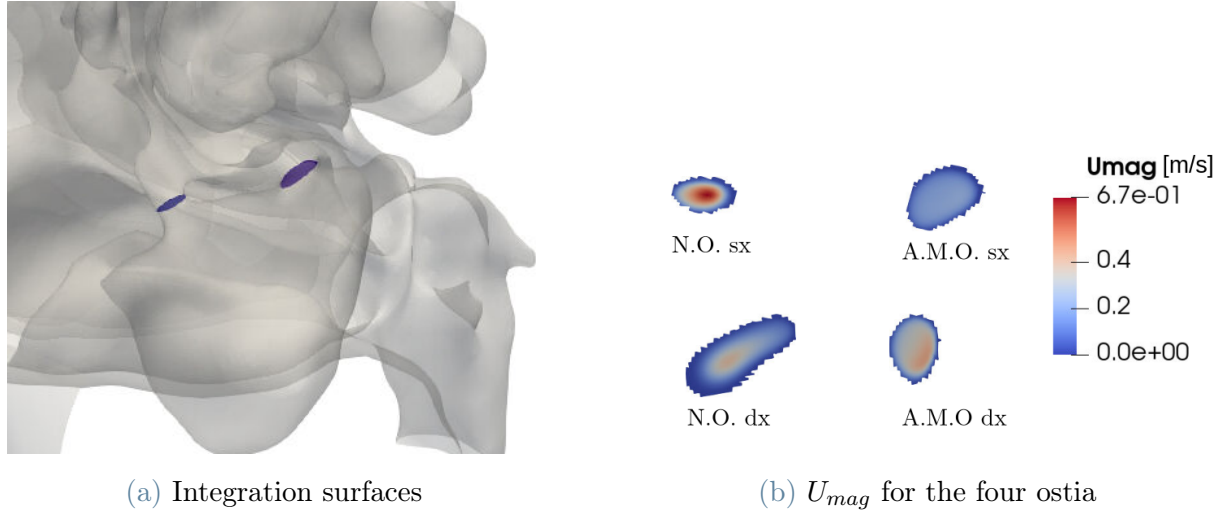


Figure 4.8: Position of the four integration surfaces for the volumetric flow through the maxillary sinuses ostia. On the right the velocity profile inside the four ostia during inspiration is reported. Above: left ostia. Below: right ostia. On the left the normal ostia (N.O.), on the right the accessory ones (A.M.O.).

In both cases, noticeable differences in the fluctuation intensity are observed inside the sinuses. In particular, panel (c) of Fig. 4.11 shows a clear increase in $\Delta\sqrt{w'^2}$, which could be interpreted as a consequence of the flow passing through the sinus in the double-ostium configuration.

A comparison between panels (a) and (b) of both figures indicates that fluctuations inside the maxillary sinus are larger in the geometry with the accessory ostium.

In a similar way, image (c) of Fig. 4.10 shows localized differences in the horizontal fluctuation component close to the ostia and within the sinus cavity. These differences suggest the presence of local recirculation structures induced by the additional opening.

Additional differences, although not unexpected, are observed inside the maxillary sinuses themselves (see Fig. 4.9(c) and (f)), highlighting the presence of a mean flow within the cavities. To better visualize the flow inside the sinus, an oblique section is used. Its position is shown in Fig. 4.12. For brevity, only the right maxillary sinus inspiration case is reported.

Recalling from Table 4.5a that the mean flow rate through the right ostium during inspiration is 2.16 mL/s, it is possible to see in Fig. 4.13(b) a jet entering the sinus with a mean velocity of approximately 0.2 m/s. On the other hand, in the physiologically normal geometry the velocity in the ostiomeatal region is higher, reaching values up to 0.68 m/s, which is twice the velocity observed in the same region for the modified geometry, where

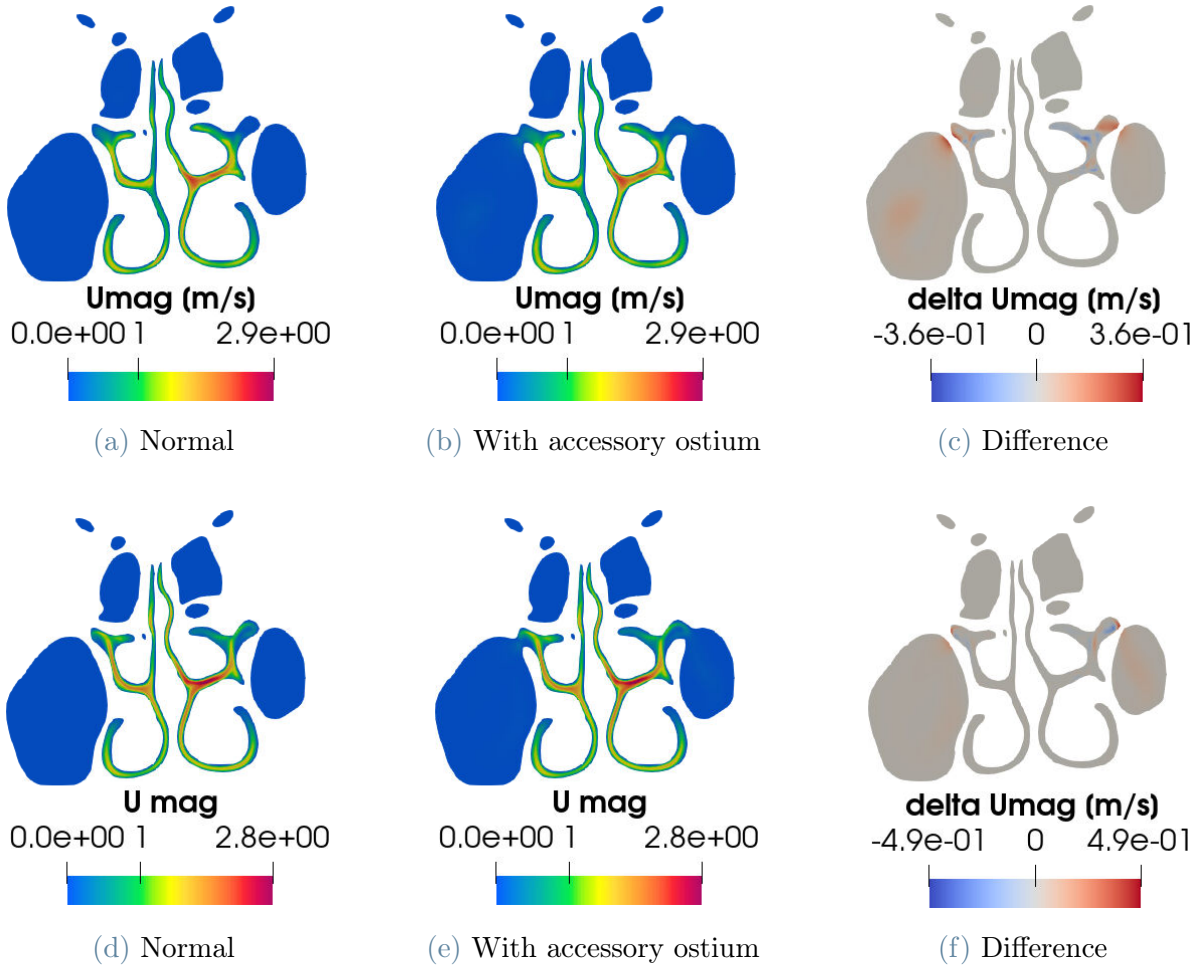


Figure 4.9: Section S3 showing the time-averaged velocity magnitude U_{mag} for the normal geometry, the geometry with the accessory ostium, and their difference. Top row: inspiration. Bottom row: expiration.

values of approximately 0.34 m/s are found.

An additional visualization of the flow in the same region is presented in the form of streamlines (Fig. 4.14). The streamlines were computed in ParaView by integrating the time-averaged velocity field using the built-in adaptive Runge-Kutta 4-5 integrator in the forward direction. The seed was defined as a sphere located inside the right normal ostium for the inspiration and the same sphere positioned in the right accessory ostium for expiration .

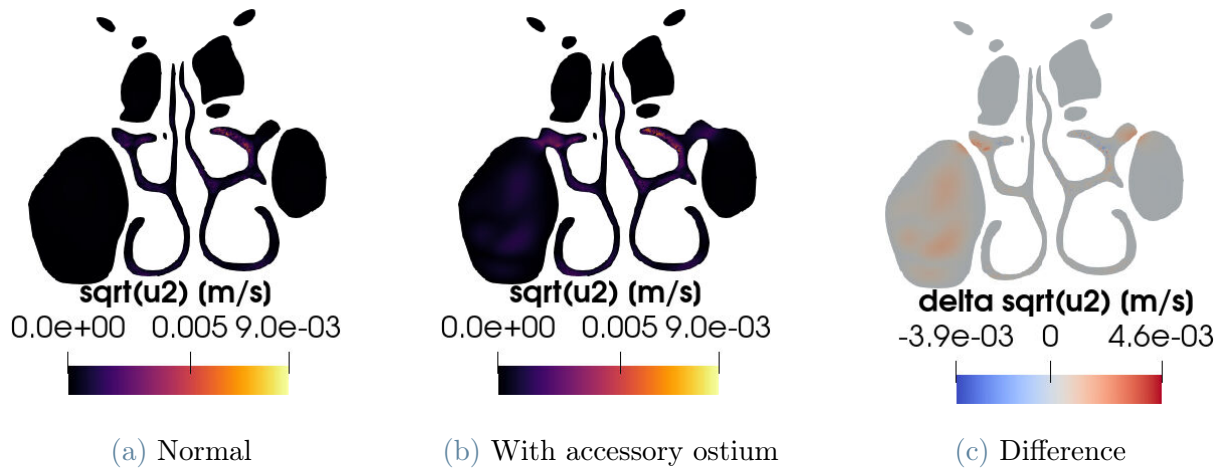


Figure 4.10: Root mean square of the velocity fluctuations in $\hat{i}$ direction, $\sqrt{u'^2}$. Section 2, with and without accessory ostium, inspiration simulation.

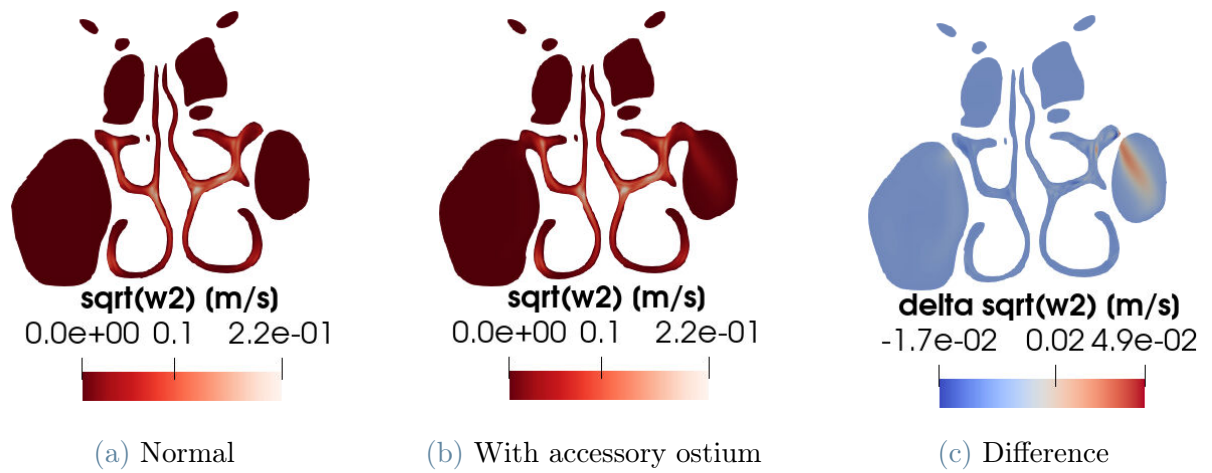


Figure 4.11: Root mean square of the velocity fluctuations in $\hat{k}$ direction, $\sqrt{w'^2}$. Section 2, with and without accessory ostium, expiration simulation.

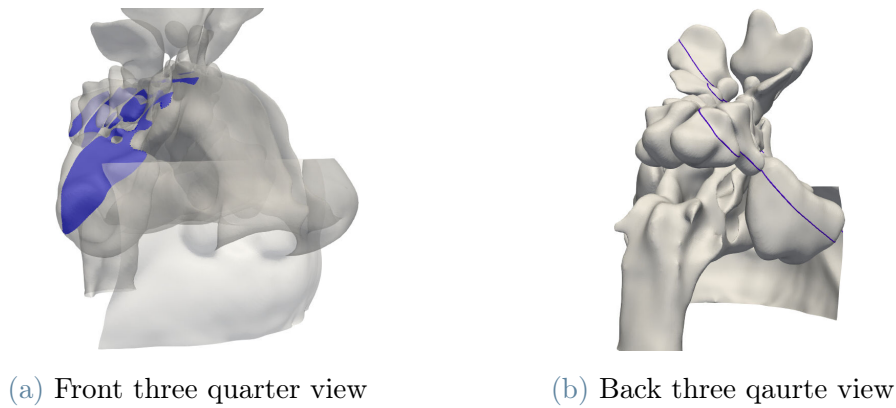


Figure 4.12: Oblique section of the right maxillary sinus. The section intersects both the normal and the accessory ostia.

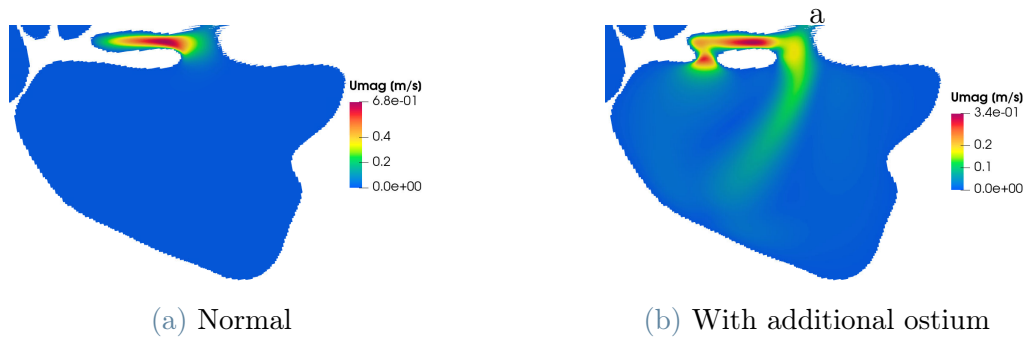


Figure 4.13: Oblique section of the right maxillary sinus for the normal geometry and the one with the AMO. Time-averaged velocity magnitude U_{mag} for the inspiration case

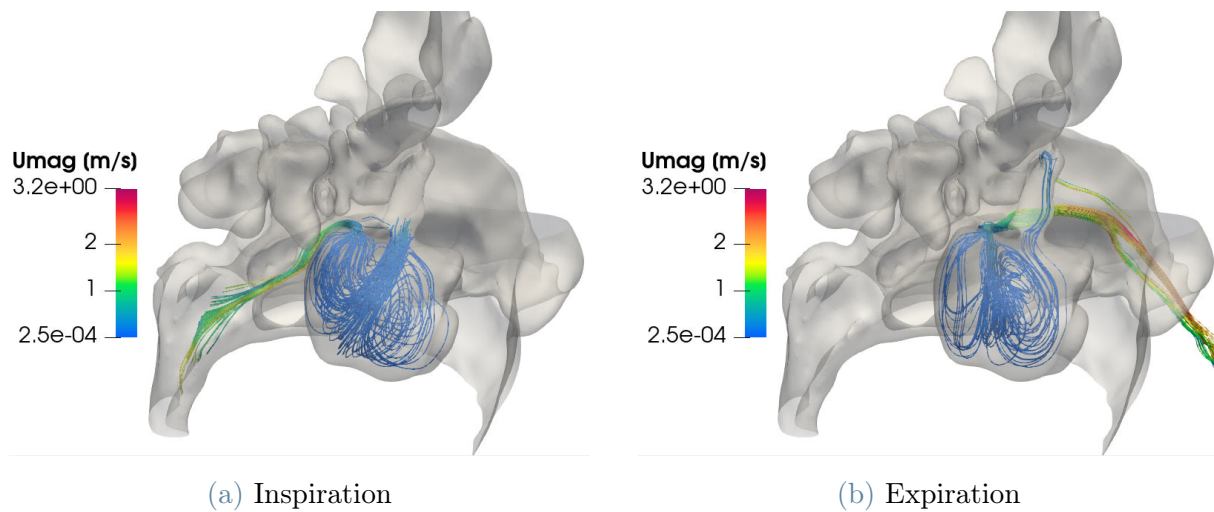


Figure 4.14: Streamlines of velocity for the inspiration (a) and expiration (b) simulations inside the right sinus for a double ostia case.

5 | Conclusions and future developments

The work presented in this thesis provides a reliable and repeatable procedure to generate computational geometries starting from CT scans. The resulting 3D models are suitable for CFD simulations and sufficiently simplified to allow the application of functional mapping projection, requirement needed in the Open Nose framework.

It has been verified that the removal of the paranasal sinuses does not fundamentally modifies the overall functionality of the nose. In particular, the net volumetric flow rate and the nasal resistance remain essentially unchanged. Even the presence of a secondary (accessory) ostium does not modify these quantities. As expected, local discrepancies and differences can be observed in regions close to the geometric modifications. However, their impact is confined to a very small area and does not significantly alter the global flow behaviour.

Therefore, it is possible to conclude that removing the sinuses and the related anatomical structures does not compromise the accuracy of the flow representation, making the use of these simplified geometries justified and appropriate.

Future research could investigate further the fluid-dynamic impact of some parameters chosen in the reconstruction procedure described in Chapter 2. For example, a CFD-based assessment could be carried out to quantify the effect of different interpolators (e.g. linear versus b-spline interpolation). A more in-depth analysis could also examine the influence of the spacing scale, particularly in upscaling, as well as on the role of the final surface smoothing factor. Part of this research is already being carried out by Dr. J. Ormiskangas at the University of Tampere, in collaboration with the Politecnico di Milano.

Finally, since the reconstruction process is time-consuming, especially when dozens or hundreds of geometries are required to expand the training dataset, it would be very beneficial to develop a ML-aided reconstruction procedure. This approach could help in the automatization of parts of the workflow, in line with some recent works such as [7, 33, 39].

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A | Appendix A - Defects

manufacturing

As discussed in Section 1.3, a key step in building the training database for the Machine Learning model is the manual modeling of anatomical defects. A total of 15 unique defects constitute the *nose vectorial base*, combining them in various ways it is possible to obtain a wide range of pathological geometries. These defects are subdivided into two main categories: septal variations and turbinate hypertrophies. Each defect is considered at two levels of severity, resulting in a total of 30 pathological configurations.

All the considered pathologies are summarized in Figure A.5, where each leaf represents a localized anatomical defect. The tree is complete, meaning that a corresponding 3D model exists for each leaf. Black leaves indicate defects that were already present in the database and were either left unchanged or modified only in a minimal way. Green leaves represent defects that were newly modelled and added during this work.

A criterion based on anatomical landmarks was introduced to define the head, body, and tail of the inferior turbinate. Specifically, the head of the inferior turbinate starts at its beginning and extends posteriorly up to just after the nasolacrimal duct. From a lateral perspective, the head terminates at the highest point of the inferior meatus. The body begins where the head ends and extends posteriorly a little bit after the maxillary sinus starts to decrease in cross-sectional area. Figure A.2 illustrates this subdivision both on a axial CT scan and on a lateral view of the corresponding three-dimensional model.

As mentioned above, each pathological condition is applied with two different severities, referred to as severity 1 and severity 2, defined as follows:

1. Partial reduction of the nasal airway, with no contact between opposing surfaces;
2. Presence of contact between opposing surfaces, resulting in partial obstruction of the airway.

Figure A.1 shows representative examples of inferior turbinate hypertrophy at the two severity levels comparing them with the corresponding healthy geometry.

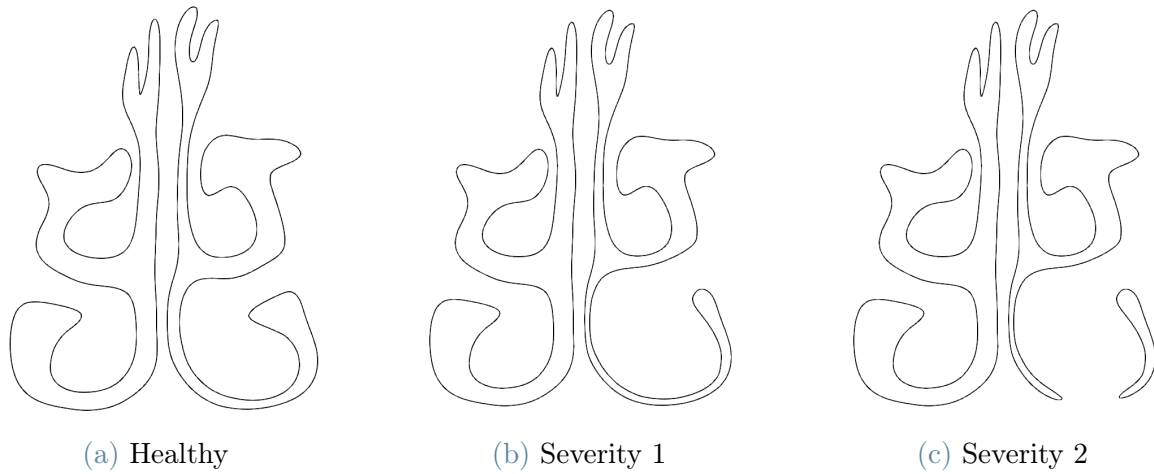


Figure A.1: Coronal section of Patient 130 showing the healthy geometry and the two hypertrophy severity. Hypertrophic body of the left inferior turbinate.

All hypertrophies were conventionally applied to the **left nasal fossa** (considering the patient's point of view). Likewise, all septal deviations were imposed as **leftward**, resulting in a reduction of the cross-sectional area of the left nasal cavity.

A brief description of the techniques used to generate the defects is provided below. The procedure described here is not mandatory, but it is the result of extensive manual experience. Two different software were employed to introduce the defects: Blender for septal deviations and septal crests, and 3D Slicer for turbinate hypertrophies. While the first choice is quite obvious given the nature of Blender, the latter is surely less conventional.

Although 3D Slicer is not designed to modify three-dimensional surface models, it was deemed particularly suitable for the creation of turbinate hypertrophies by manually painting additional voxels within the nasal cavity. The ability to paint directly on coronal CT slices and have an effect on the 3D surface was proved extremely useful for introducing anatomically realistic hypertrophies. This approach also allowed to see in real-time the fraction of the nasal fossa obstructed by the added tissue.

Figure A.3 shows the mask (in green) applied to the CT scan to select the regions that will be converted into the three-dimensional model. Although this method is highly accurate from an anatomical point of view, it is also extremely time-consuming. Each slice must be manually painted, and the total number of slices can grow significantly. For example, creating a 5 cm long hypertrophy requires painting approximately 100 slices if the scan has a spatial resolution of 0.5 mm. This added complexity is though compensated by the high anatomical fidelity of the resulting defect. Figure A.1 shows the result of the

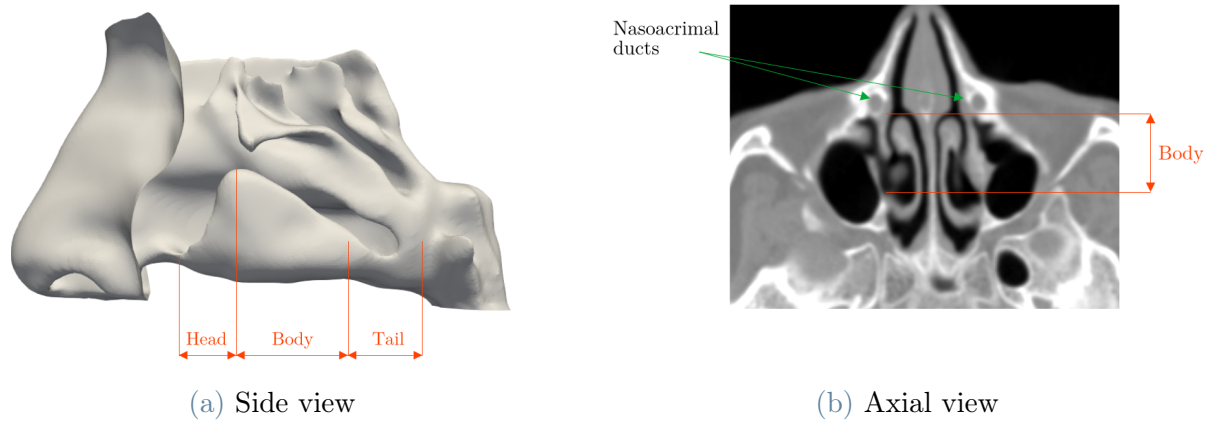


Figure A.2: Side view and axial section of patient 130. Subdivision of the inferior turbinate.

insertion of the pathology shown in Figure A.3.

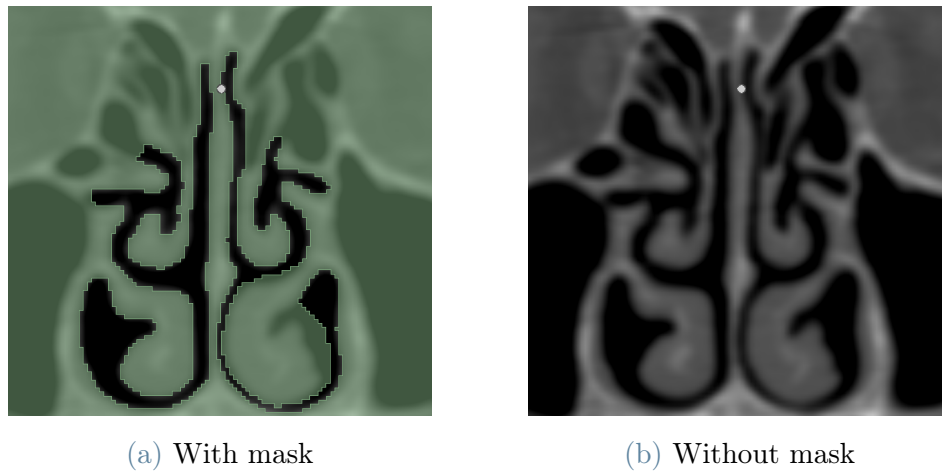


Figure A.3: Coronal section of patient 130 showing the same slice with and without the masking segment that is then converted in 3D geometry.

Blender was instead considered more suitable for the introduction of septal deviations and septal crests. All modifications were performed within Blender's sculpting environment. The main brushes used were *Grab*, *Grab 2D*, *Grab Silhouette*, and *Inflate/Deflate*. Even if they appear similar, each tool serves a specific purpose and enables precise control over localized geometric deformations.

When introducing septal deviations, it was carefully considered the possible presence of compensatory hypertrophy on the opposite side. Figure A.4 shows a comparison between a healthy geometry and a model with an added septal deviation in the left posterior-inferior

region.

Since three-dimensional visualizations is quite an unusual representation for clinicians, a small Python script was developed to slice the STL geometry along the three anatomical planes (coronal, axial, and sagittal). This allows the modified surface model to be visualized in a CT scan fashion. This enables a faster medical review by presenting the defects in a way doctors are more used to.

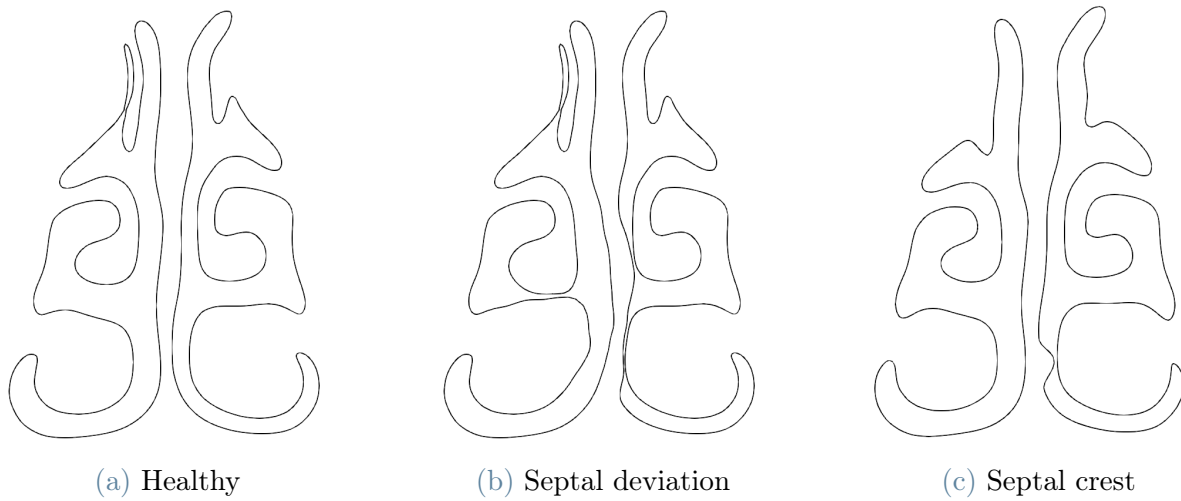


Figure A.4: Coronal section showing the healthy geometry and the two added septal variations (severity 1) on Patient 130. Both defects are located in the inferior posterior area.

Given the large number of defects and associated files, a consistent naming convention was introduced to ensure clarity. Each defect is labeled according to the following scheme:

Pathology type + Macro area + Position + severity

The following abbreviations are proposed:

- **HY:** Hypertrophy
- **SD:** Septal Deviation
- **SC:** Septal Crest

A complete list of the proposed names is reported in table A.1.

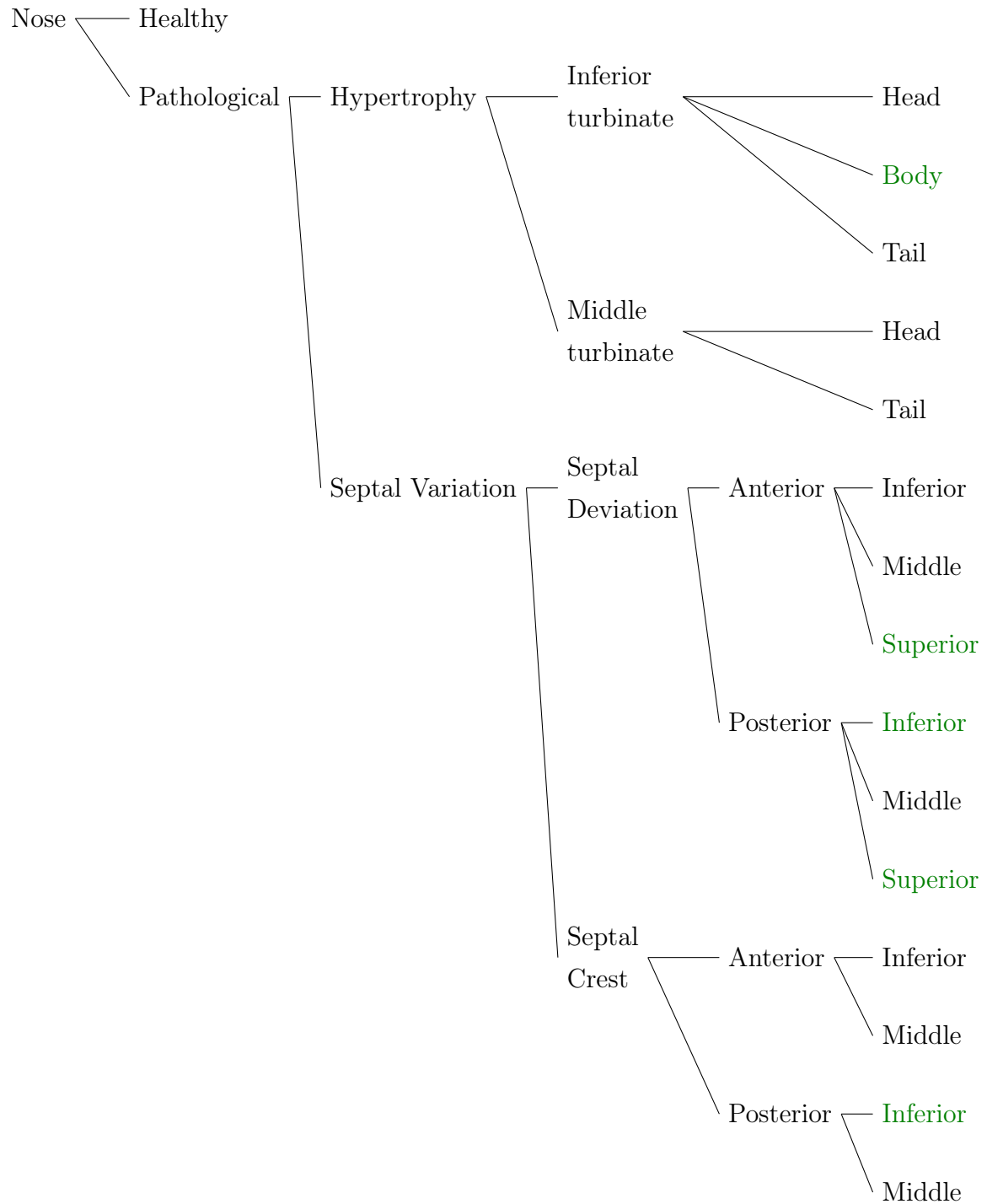


Figure A.5: Pathologies tree. All leaves are existing 3D models. Green leaves represent the new additions made during this work

Pathology	Macro area	Position	Name
Hypertrophy	Inferior turbinate	Head	HY_inf_head
		Body	HY_inf_body
		Tail	HY_inf_tail
	Middle turbinate	Head	HY_mid_head
		Tail	HY_mid_tail
Septal Deviation	Anterior	Inferior	SD_ant_inf
		Middle	SD_ant_mid
		Superior	SD_ant_sup
	Posterior	Inferior	SD_post_inf
		Middle	SD_post_mid
		Superior	SD_post_sup
Septal Crest	Anterior	Inferior	SC_ant_inf
		Middle	SC_ant_mid
	Posterior	Inferior	SC_post_inf
		Middle	SC_post_mid

Table A.1: Proposed naming convention for the defects.

B | Appendix B - Additional images

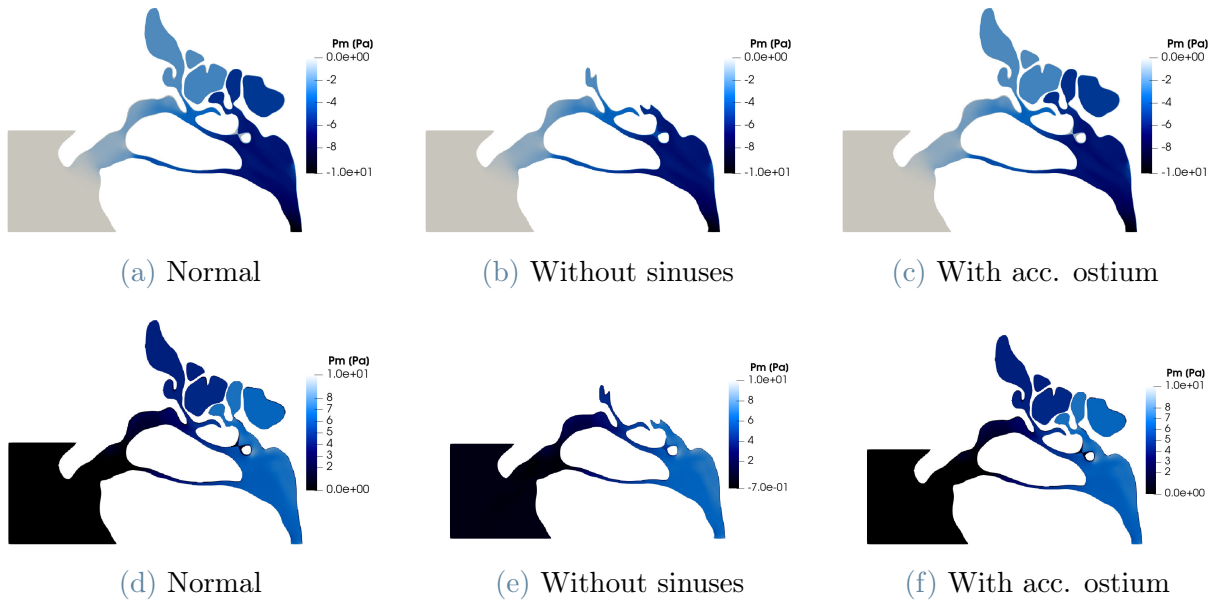


Figure B.1: Time-averaged Pressure (P_m) [Pa] in sagittal section. Top row: inspiration. Bottom row: expiration. The three configurations show almost identical pressure distributions.

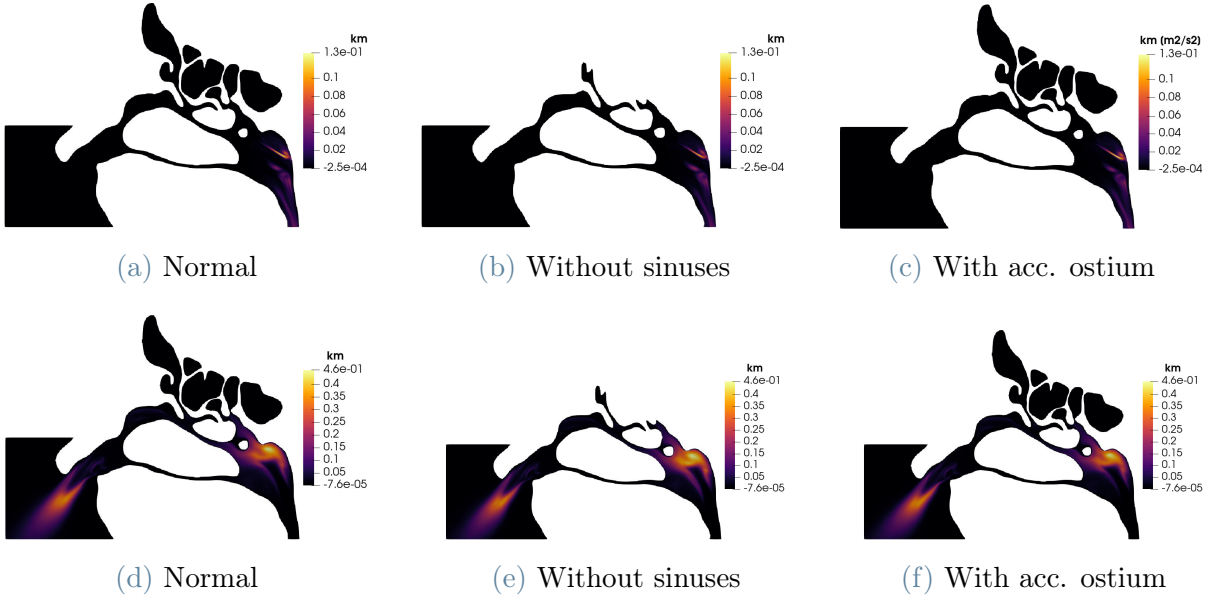


Figure B.2: Turbulent kinetic energy k [m^2/s^2] in sagittal section. Top row: inspiration. Bottom row: expiration. The three configurations show almost identical distributions.

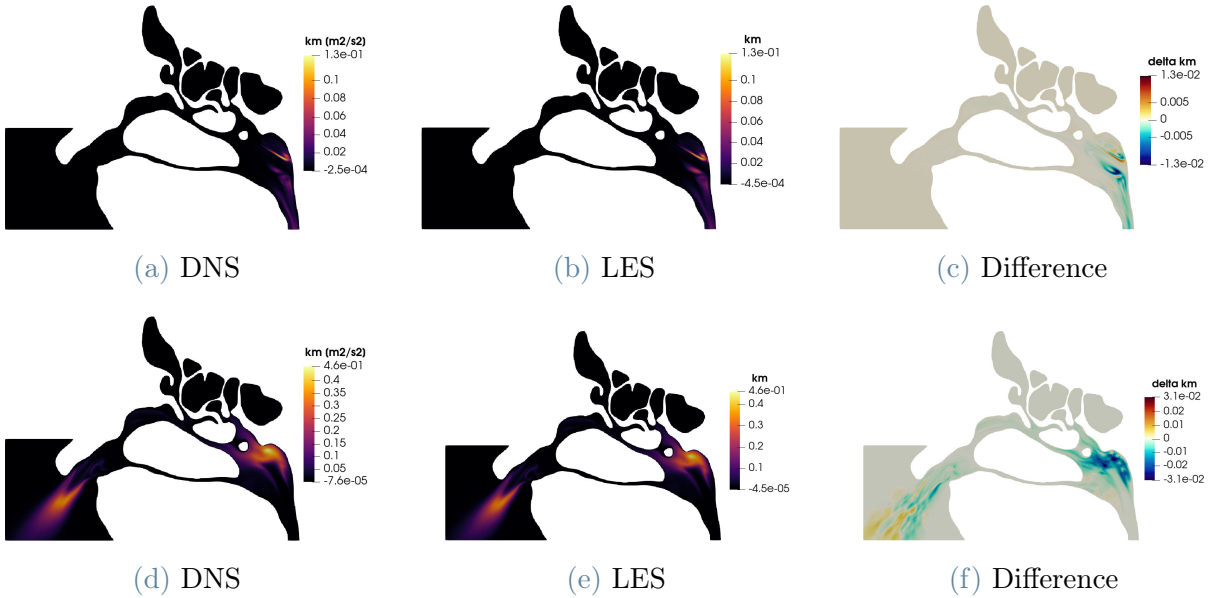


Figure B.3: Turbulent kinetic energy k [m^2/s^2] in sagittal section. Top row: inspiration. Bottom row: expiration. DNS and LES.

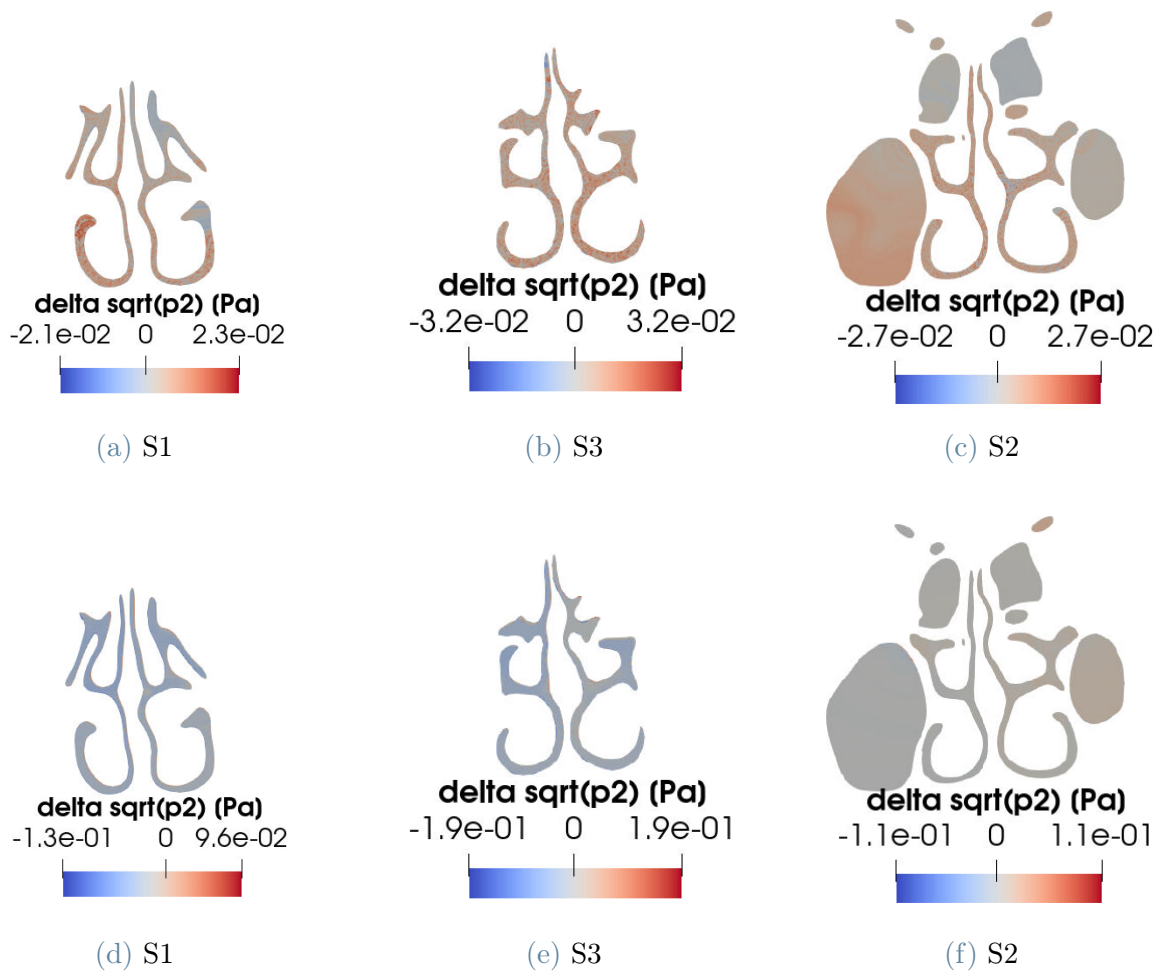


Figure B.4: Difference in the root mean square of the pressure oscillation $\overline{p'^2}$ between with sinuses and not (a, b, d, e) and with and without accessory ostia (c, f). Top row: inspiration. Bottom row: expiration.

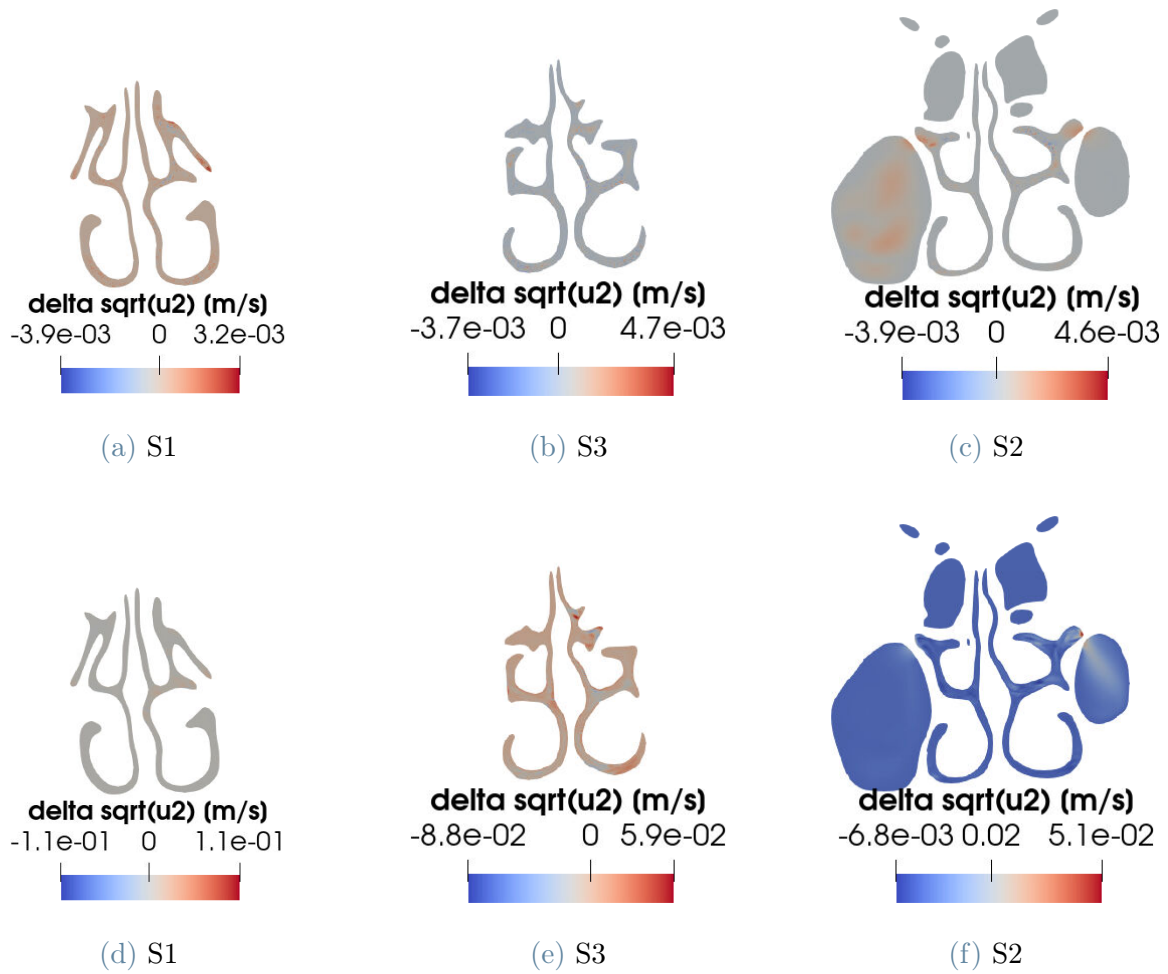


Figure B.5: Difference in root mean square velocity oscillation $\overline{u'^2}$ between with sinuses and not (a, b, d, e) and with and without accessory ostia (c, f). Top row: inspiration. Bottom row: expiration.

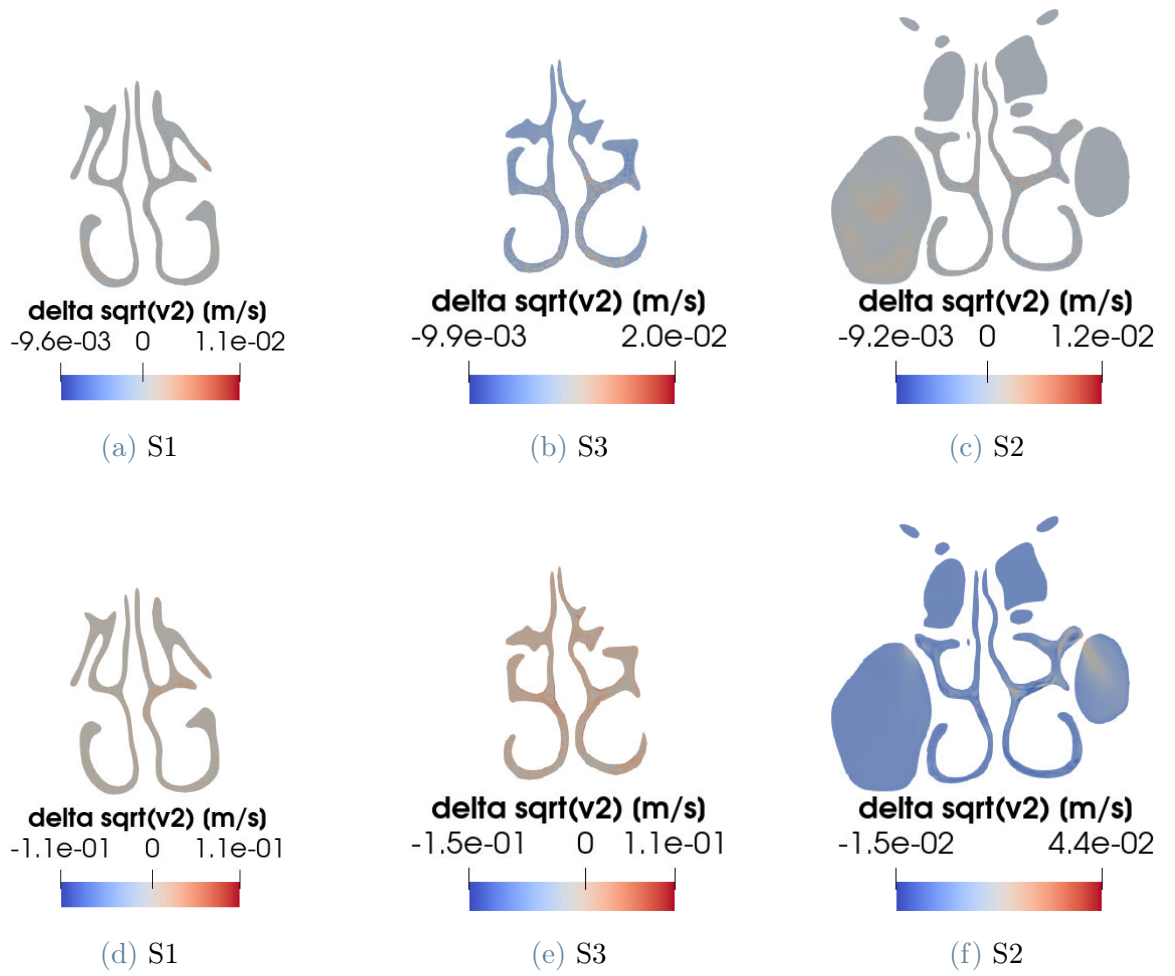


Figure B.6: Difference in the root mean square velocity oscillation $\overline{v'^2}$ between with sinuses and not (a, b, d, e) and with and without accessory ostia (c, f). Top row: inspiration. Bottom row: expiration.

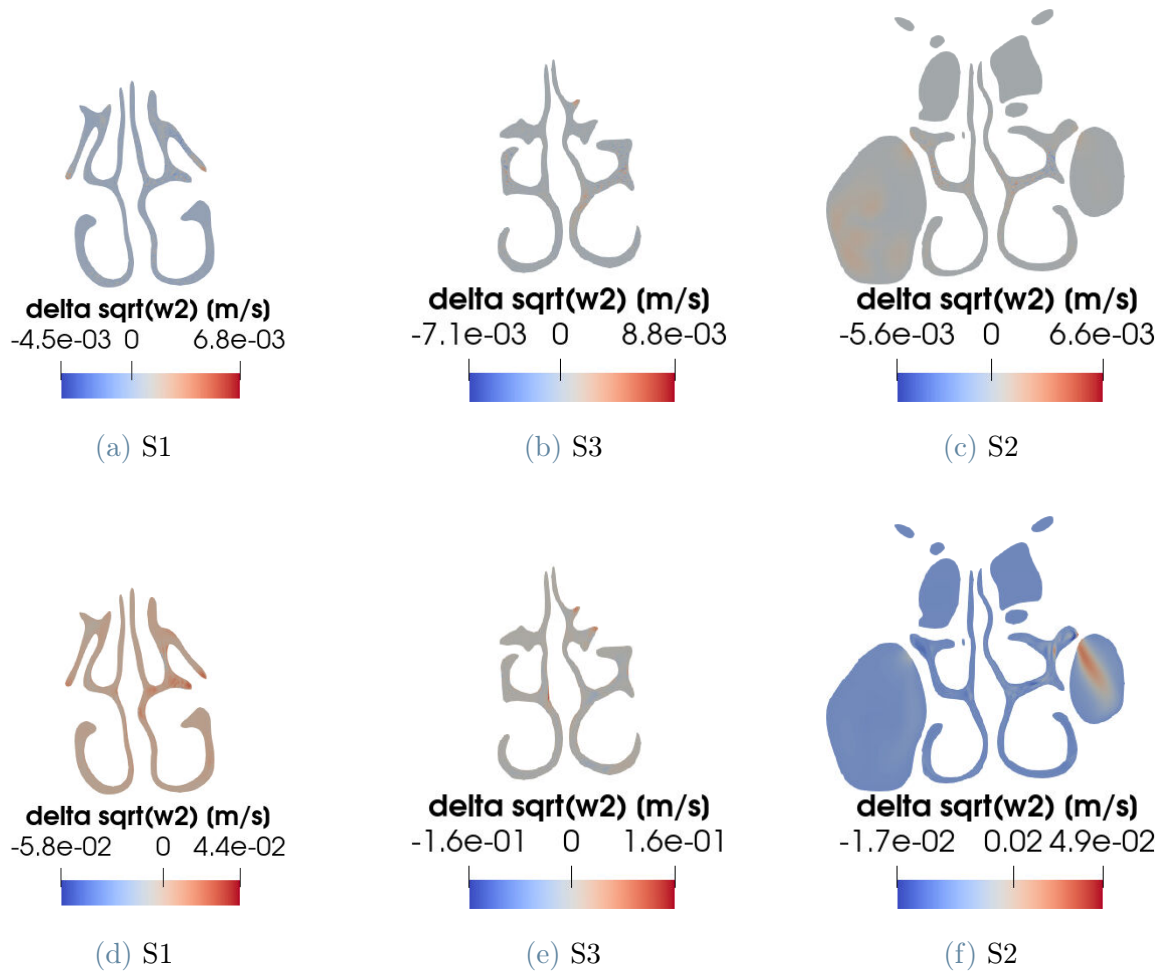


Figure B.7: Difference in root mean square velocity oscillation $\overline{w'^2}$ between with sinuses and not (a, b, d, e) and with and without accessory ostia (c, f). Top row: inspiration. Bottom row: expiration.

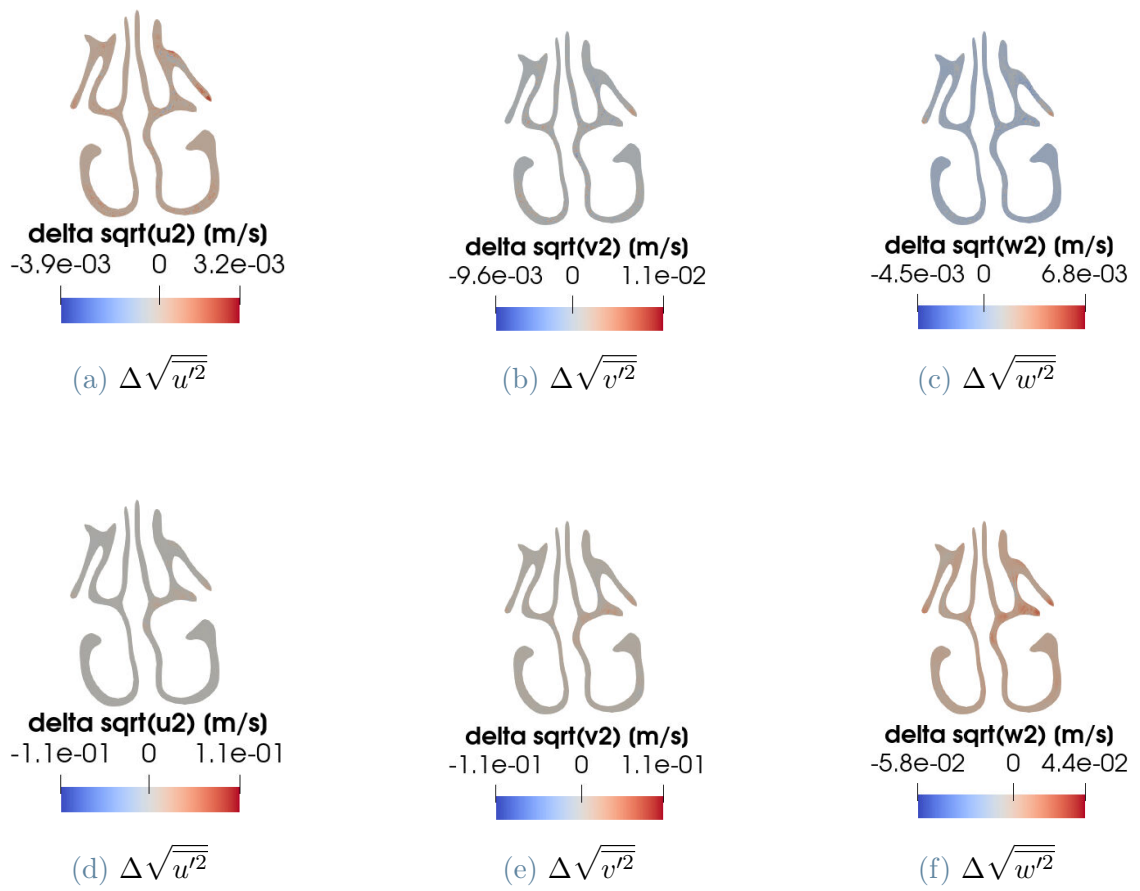


Figure B.8: Difference in root mean square of velocity oscillation $\overline{u'^2}$, $\overline{v'^2}$, $\overline{w'^2}$ between with and without sinusoidal oscillations in section S1. Top row: inspiration. Bottom row: expiration.

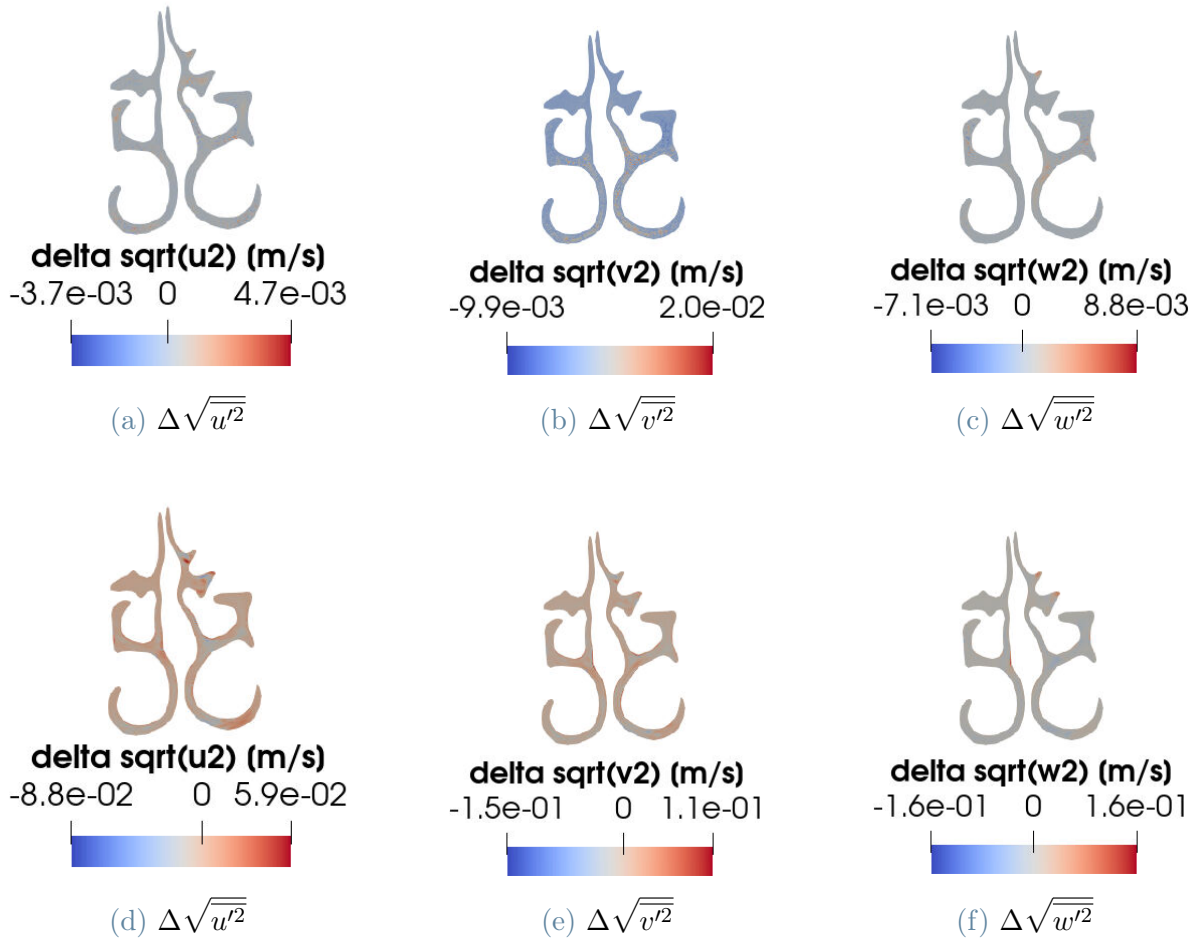


Figure B.9: Section S3. Difference in root mean square of velocity oscillation in the three directions $\overline{u'^2}$, $\overline{v'^2}$, $\overline{w'^2}$ between with sinus and not in section S3. Top row: inspiration. Bottom row: expiration.

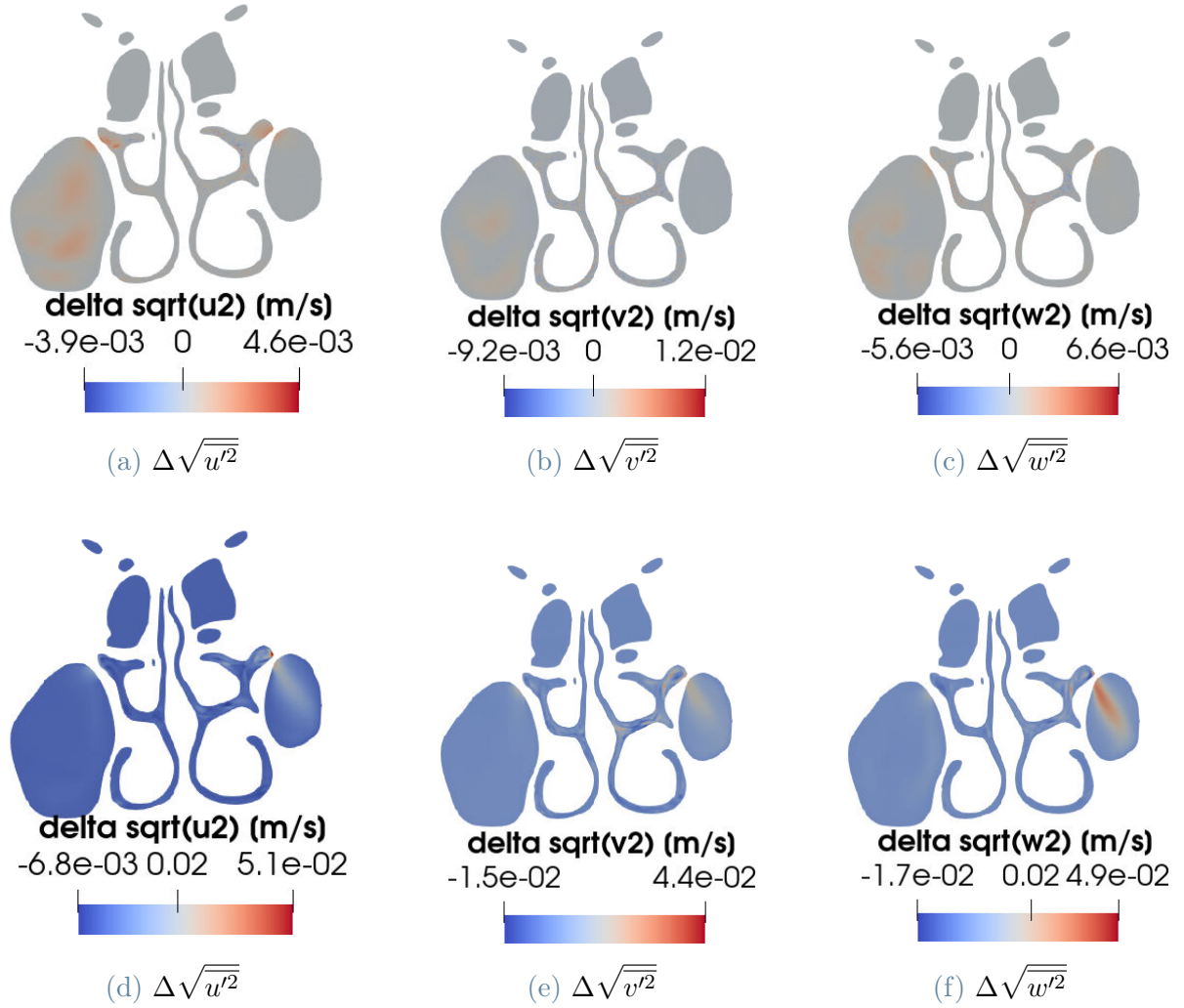


Figure B.10: Difference in root mean square of velocity oscillation in the three directions $\overline{u'^2}$, $\overline{v'^2}$, $\overline{w'^2}$ between with and without accessory ostia, in section S2. Top row: inspiration. Bottom row: expiration.

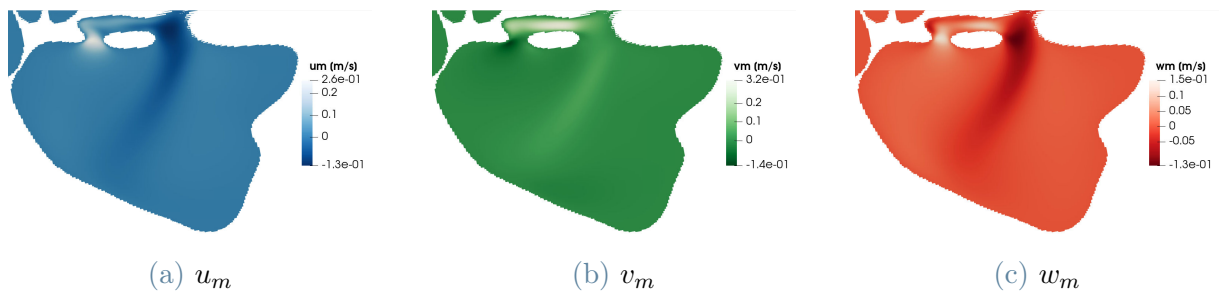


Figure B.11: Oblique section of the right maxillary sinus of the complete geometry (with acc. ostium), showing the timed-averaged velocity components u_m , v_m and w_m during inspiration simulations

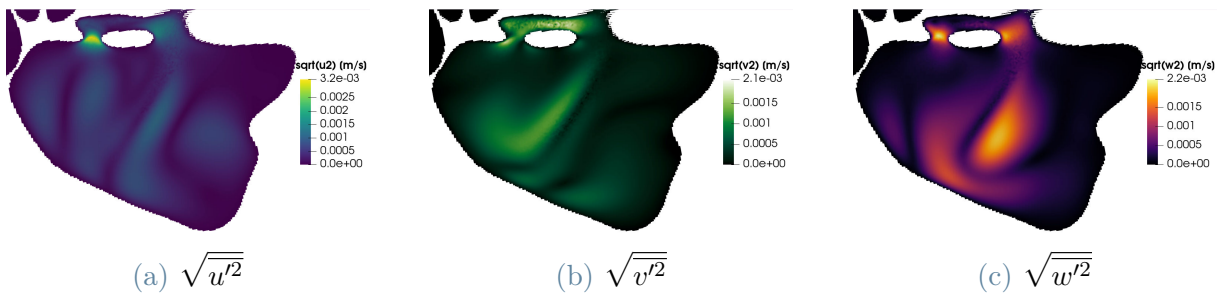


Figure B.12: Oblique section of the right maxillary sinus of the complete geometry (with acc. ostium), showing the root mean square of the velocity oscillations components $\sqrt{u'^2}$, $\sqrt{v'^2}$ and $\sqrt{w'^2}$ during inspiration simulations.

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Acknowledgments

I would like to express my sincere gratitude to Professor Quadrio for giving me the opportunity to take part in the OpenNose project, contributing to it through this thesis, and for his dedicated supervision throughout its entire development.

I would also like to thank Dr. Marco Atzori and Dr. Jaakko Ormiskangas for their technical and scientific support during the research activity and the analysis of the results.

Finally, I would like to thank the medical staff of San Paolo Hospital for the time and attention they devoted to the 3D modeling phase of this work.

