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Brigham Young University

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Wall Shear Stress in Simplified and Scanned Avian Respiratory Airways

Michael Sterling Farnsworth

A thesis submitted to the faculty of
Brigham Young University
in partial fulfillment of the requirements for the degree of
Master of Science

Scott L. Thomson, Chair
Steven E. Gorrell
Bradley R. Adams

Department of Mechanical Engineering
Brigham Young University

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ABSTRACT

Wall Shear Stress in Simplified and Scanned Avian Respiratory Systems

Michael Sterling Farnsworth
Department of Mechanical Engineering, BYU
Master of Science

Birds uniquely produce sound through a vocal organ known as a syrinx. The presence of wall shear stress acting on the airway cells of any organism will affect how airway cells develop and multiply. Unique features of avian airway geometry and breathing pattern might have contributed to the development of the syrinx. This thesis examines wall shear stress in the trachea and first bronchi of avian geometries using computational fluid dynamics. The computational fluid dynamic simulations underwent grid- and time-independence studies and were validated using particle image velocimetry. Parameters such as bird size, bronchial branching angle, and breathing waveform were examined to determine conditions that contributed to higher wall shear stress. Both simplified and CT scan-derived respiratory geometries were examined. Maximum wall shear stress for the simplified geometries was found to be highest during the inspiratory phase of breathing and was highest near the pessulus. Maximum wall shear stress in the CT scan-derived geometries was less phase-dependent and was highest near constrictions in the airway. Comparison between scanned and simplified geometry simulations revealed significant differences in wall shear stress magnitudes and flow features. If wall shear stress is found to be important in the development of the syrinx or the advantage of a syrinx, the thesis results are anticipated to aid in characterizing conditions that would have contributed to the development of the syrinx or advantages of syringeal vocal fold position over tracheal vocal fold position.

Keywords: wall shear stress, CAT scan, respiratory system, bird, avian, CFD, PIV
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I’d like to thank my committee members. I greatly respect both Dr. Adams and Dr. Gorrell and have enjoyed their classes and have appreciated their advice and direction. I am also, of course, very thankful for everything that Dr. Thomson has done on my behalf. I’ve always felt that any advice he has given has been with my best interests in mind. I appreciate his patience and goodness.

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1 INTRODUCTION

The purpose of this thesis is to explore wall shear stress in the avian respiratory system using computational fluid dynamics simulations. This chapter outlines the motivation for doing so, as well as summarizes previous studies to give context for the research presented in this thesis. Contributions are outlined and a short summary of the remainder of the thesis is presented.

1.1 Avian Vocal Production

Humans vocalize using flow-induced vibration of the vocal folds, located in the larynx near the top of the trachea. Similar to humans, birds use vibrating tissue to vocalize [1]. However, in contrast to other tetrapods (amphibians, reptiles, and mammals), birds vocalize with an organ called the syrinx. The syrinx is a novel vocal organ [2] which is located near the tracheobronchial junction (the location where the trachea and bronchi intersect, abbreviated TBJ). The precise location of the syrinx varies between species, and syrinx classification is dependent on whether it is above the TBJ (tracheal syrinx), in the TBJ (tracheobronchial syrinx), or below the TBJ (bronchial syrinx) [1]. Illustrations of these types of syrinx configurations are shown in Fig. 1.1. It is not generally understood what natural advantage might have contributed to the development of this organ, or why so much variation exists.
1.2 Avian Respiratory System Studies

It has been known since the 1970’s that, unlike other animals with lungs, airflow within birds’ lungs is unidirectional [3]. Recent findings have revealed that this trait is also shared among other archosaurs (crocodiles and alligators), and has also been discovered in two squamate lizards, a monitor lizard and an iguana [3,4]. This was considered significant because the presence of unidirectional flow in non-flying bird relatives suggests that unidirectional flow may not have developed to effectively meet the oxygen demands of flying, as had been previously thought [3,4]. These studies have relied on in vivo measurements using flowmeters as a method of data quantification. Interest in this unique feature of unidirectional flow has given rise to computational fluid dynamics studies on the respiratory system of birds in recent years [5,6]. One such study created a simplified model of the respiratory system of an ostrich in order to model the aerodynamic valving that was thought to play a part in the unidirectional behavior of the airflow in the avian respiratory system [5]. This study used several geometric models with varying levels of completeness, and found that aerodynamic valving did occur in the more complete models, but

Figure 1.1: Depictions of a tracheobronchial syrinx (left), a tracheal syrinx (center), and a bronchial syrinx (right). The LL (later labia), ML (medial labia), MTM (medial tympaniform membrane), and LTM (lateral tympaniform membrane) are the structures involved in sound production. Adapted from Mindlin and Laje (2005). Used with permission.
not the partial models that were in some way truncated [5]. Another study varied the compliance (which they defined as the change in volume with pressure) of a simplified domestic fowl airway on the resulting flow patterns, and found that varying compliance could change whether or not the flow was unidirectional [6]. Other avian respiratory studies have used computational fluid dynamics in order to study heat transfer and water loss in bird breathing cycles, with the intent of extrapolating results to now extinct animals, and concluded that the domestic fowl likely loses between 13-19% of its starvation metabolic rate to heat loss via breathing [7,8]. However, up until the present, avian computational fluid dynamics studies have not focused on wall shear stress or the respiratory system near the TBJ with a focus on the syrinx.

1.3 Wall Shear Stress

Wall shear stress (WSS) is the force per unit area that a fluid moving relative to a surface imparts on the surface. WSS has been studied in correlation with cell damage in the human respiratory system [9-14], as high levels of WSS can cause cellular damage [9-17]. Any stresses, including WSS, can change the way that cells reproduce, grow, and adapt to their environment [18]. Compliant surfaces result in significantly lower magnitudes of WSS, with one study finding a fivefold reduction of WSS in a flexible model compared to a rigid model [11]. Having a compliant region, namely the syrinx, at a region of high WSS might therefore be advantageous by possibly reducing the WSS experienced by cells in the region that would otherwise experience elevated WSS levels. Showing whether or not the TBJ experiences higher WSS due to breathing patterns coupled with a study of associated mechanobiology may shed light on whether WSS might help explain the presence of the syrinx in birds.
1.4 **Respiratory System Studies**

While studies do exist involving the respiratory systems of birds (see Section 1.2), most respiratory studies have involved the human respiratory system. While the respiratory system of birds is unique in many ways, there are also many similarities to humans, and understanding the background and conventions in human respiratory studies can lay the foundation for approaches of the present work. The human respiratory system features bidirectional internal flow with time-varying boundary conditions, which is different than many traditional CFD applications, and requires unique treatment.

1.4.1 **Non-Dimensional Numbers in Respiratory Studies**

Two key non-dimensional numbers that are used in respiratory studies are the Reynolds number (Re) and the Womersley number (Wo) [5, 6, 12, 18, 19]. The Reynolds number, commonly used in many fluids applications, represents a ratio of inertial to viscous forces and is used to determine whether a flow is laminar or turbulent. Internal flows with Re < 2100 are generally considered to be laminar [20]. The Reynolds number is defined as

$$Re = \frac{\rho V D}{\mu}$$  \hspace{1cm} (1.1)

where $\rho$ represents the density of the fluid, $V$ represents the average velocity of the fluid, $D$ represents some important length scale (generally the diameter or hydraulic diameter for internal flows), and $\mu$ represents the dynamic viscosity of the fluid.

The Womersley number, commonly used in biofluid studies, gives a ratio of pulsatile flow frequency to viscous forces, and predicts whether or not a flow can be treated as quasi-steady [19]. Flows with Wo < 1 are treated as quasi-steady, while flows with Wo > 1 are treated as unsteady.
[19], although some studies have suggested that Wo as high as 4 and 5 could be used to justify a quasi-steady approach [19, 21]. The Womersley number is defined as

\[ Wo = L \left( \frac{\omega}{\nu} \right)^{\frac{1}{2}} \]

where \( L \) refers to a characteristic length (the radius for internal flows), \( \omega \) refers to the frequency of oscillation, and \( \nu \) is the kinematic viscosity.

For internal flows, both the Reynolds number and the Womersley number affect the flow patterns that are expected to occur, as well as the WSS magnitude. In a straight pipe of a given diameter, assuming that the fluid does not change properties, a higher Re number will increase the WSS due to steeper velocity gradients near the wall. Whether a higher Wo number results in higher WSS depends on some of the underlying assumptions. If flow rates are equal, WSS will be greater at higher Wo numbers since the velocity profile becomes more plug-like at higher Wo numbers, resulting in a steeper gradient at the wall. Conversely, if the flow is pressure-driven and the maximum pressure magnitude does not change, a higher Wo number will result in lower WSS. This is because despite a change in profile shape, the WSS decreases due to a decrease in flow rate [19]. This has been quantified mathematically as a magnitude drop of 8% at Wo = 1 and a drop of 50% near Wo = 2 [19]. In airways of birds of different sizes, frequencies, velocities, and diameters (and thus Re and Wo values) all simultaneously change, which makes predicting changes in WSS nontrivial, and warrants the use of computational approaches.

Previous CFD studies in humans have explored, for example, the effect of changing the breathing profile of a high frequency oscillatory ventilator (HFOV) to resulting WSS, as HFOVs have been linked to cellular damage [9, 12]. However, to the best of the author’s knowledge,
studies involving WSS in the respiratory system of birds have never been conducted. Additionally, studies involving birds have not considered various breathing profiles.

1.4.2 Imaging in Respiratory Studies

While early work using both experimental and computational fluid dynamics have focused on simplified models of the respiratory system, with advances in medical imagery, many recent studies of the human respiratory system have been of models generated by CT scans, or other similar techniques [10, 11, 13, 18, 22-24]. Studies on human geometries have revealed that even among humans, different morphologies in the human respiratory system have resulted in different flow patterns [22-24]. Computational studies up to the present of birds’ respiratory systems have been simplified to at least some degree, and have not incorporated scanned geometries [5-8].

1.5 Research Objectives and Contributions

This thesis is concerned with characterizing the WSS distribution in the avian respiratory system near the TBJ. To this end, several factors contributing to the overall WSS distribution and magnitude were explored. The research presented here builds upon the existing literature, while contributing the following unique aspects:

- While respiratory WSS has been explored previously in humans [9-14] and rats [18], it has not previously been explored in birds. This thesis explores WSS in birds.
- Some human respiratory studies have used time-varying boundary conditions that were not simplified and sinusoidal, but rather, derived from measured pressure or velocity measurements in patients [25]. This has not been applied to bird simulations. This thesis compares a waveform from bird measurements to simplified waveforms.
While other respiratory studies of birds have included the TBJ, the flow around this region has not been the focus of these studies [5-8]. The work in this thesis centers on the TBJ.

CT and other medical imaging scans are commonly in human respiratory studies [10, 11, 13, 18, 22-24]; bird simulations have only used simplified geometries [5-8]. This study explores the effects of simplified versus scanned geometries on WSS and flow patterns.

1.6 Thesis Overview

1.6.1 Methodology (Chapter 2)

The general methodology for computational fluid dynamics simulations of a simplified geometric representation of the avian airway is presented, including solver settings, geometric presentation, and boundary conditions. Time- and grid-independence studies are presented. Results from particle image velocimetry to validate the simulations are reported.

1.6.2 Simplified Geometry (Chapter 3)

Results from the simplified geometry studies are presented, including velocity and WSS spatial distributions, maximum and minimum WSS locations, and secondary flow patterns. The effects of bird size, breathing pattern, and bronchial branching angle on these output parameters are each individually evaluated, and the underlying causes behind observed differences are elaborated. The effect of unsteady flow is evaluated and quantified.
1.6.3 Scanned Geometry (Chapter 4)

Computational fluid dynamics simulations were performed using airway geometries obtained from CT scans of a hummingbird and a little swift. CT scan settings and the geometric preparation methods are set forth. The results of both species are presented for velocity and WSS spatial distributions, maximum and minimum WSS locations, and secondary flow patterns. The results are compared to simplified geometry results with properties based on the same mass.
2 METHODOLOGY, VERIFICATION, AND VALIDATION OF SIMPLIFIED GEOMETRY AVIAN AIRWAY SIMULATIONS

This chapter introduces the general settings and parameters used in the computational fluid dynamics simulations, with a focus on the simplified models (Chapter 3), although many settings apply to the scanned models (Chapter 4) as well. Boundary conditions are given and justified. The method and results for grid and time independence studies are described. Validation of the simulations is presented using particle image velocimetry in a complementary experimental setup.

2.1 Simplified Geometries

Simplified geometries of the avian trachea and first bronchi consisted of straight, intersecting cylinders, as depicted in Fig. 2.1. In these simplified models of the airway, the tracheal length, $L_1$, and tracheal diameter, $D_1$, were varied parametrically according to allometric relationships discussed in Sec. 2.2. The bronchial diameter, $D_2$, was the same as the tracheal diameter. Researchers have suggested a ratio of $2^{1/3}$ for diameters in successive generations in a symmetric human airway [26]. The avian airway may or may not follow this model. A depiction of a coronal section of five different bird species is shown in Fig. 2.2. While it would be premature to make a conclusive generalization of a tracheal to bronchial diameter ratio based on these images, the assumption of a 1/1 ratio in the trachea and first bronchi near the TBJ is not unreasonable. This assumption further removes the need for additional assumptions regarding how to model the
reduction in area from the trachea to the bronchi (see Fig. 2.1). Bronchial lengths, $L_2$, were modeled as 0.43 times the length of the trachea. Bronchial branching angles, $\alpha$, from $15^\circ$ to $90^\circ$ were explored.

The intersections between the trachea and the bronchi were rounded using a fillet with a radius of half of the tracheal diameter. The intersection between the two bronchi, corresponding to the pessulus (see Fig. 2.1), was also rounded. In order to approximate the pessulus radius, images of the coronal section of five different bird species near the TBJ were used (see Fig. 2.2). For each bird, the approximate radius of curvature was digitally measured using these images, as well as

Figure 2.1: Depictions of a typical simplified geometry of the avian respiratory airway with a $40^\circ$ branching angle used in the CFD simulations.
the approximate diameter of the trachea. The average ratio of tracheal diameter to pessulus radius was found to be 15.6. This value was then used to scale the fillet diameter based on the allometrically-determined tracheal diameter, discussed in Sec. 2.2. While it is certainly possible that the pessulus rounding follows an allometric relation other than that discussed in Sec. 2.2, insufficient data were available to determine the validity of a constant ratio assumption.

Figure 2.2: Coronal images used to approximate the radius of curvature of the pessulus for the simplified geometry avian airway simulations. Images courtesy Dr. Tobias Riede.
2.2 Allometric Equations

In biology, allometry refers to how organisms change with body size. Allometric equations mathematically describe relationships between biological features and body size. While the data necessary for setting up CFD simulations of simplified avian airways are available for many birds (tracheal diameters and lengths, as well as breathing rates and tidal volume), allometric equations offer the ability to describe the conditions for a bird of a general size, and can help interpolate between body sizes. In this study, allometric equations for tidal volume and breathing frequency [27] were used in every CFD simulation, while allometric equations for tracheal length and diameter [28] were used for the simplified geometries only. The allometric equations used are listed below. The units are: tidal volume [mL], mass [kg], frequency [min⁻¹], length [cm], and diameter [cm].

\[
\text{Tidal Volume} = 22.9 \cdot \text{Mass}^{1.08} \quad (2.1)
\]

\[
\text{Breathing Frequency} = 17 \cdot \text{Mass}^{-0.34} \quad (2.2)
\]

\[
\text{Tracheal Length} = 15.75 \cdot \text{Mass}^{0.384} \quad (2.3)
\]

\[
\text{Tracheal Diameter} = 0.533 \cdot \text{Mass}^{0.337} \quad (2.4)
\]

2.2.1 Mass of Birds

There are thought to be approximately 18,000 species of extant birds [29]. The smallest known bird is the 2 gram bee hummingbird, *Calypte helenae*, while the largest extant bird is the ostrich, *Struthio camelus*, which weighs about 80 kg [30]. Bird masses of 3 g, 30 g, 300 g, 3,000 g, and 30,000 g were considered in the present work, which encompass nearly all extant birds [30]. Additionally, peak flow values for birds larger than about 80 kg begin to transition into the
turbulent regime (Re > 2100) [20]. While turbulent-level simulations could potentially change the trends and results obtained, undertaking the investigation of turbulent-level simulations was left to future work. As approximately 80% of birds studied in [30] fall within the 3 to 300 g range, in simplified geometry simulations where mass was not varied, a 30 g bird size was used.

2.3 Boundary Conditions

Boundary conditions are critical for obtaining simulations that are true to life. Many respiratory simulations have used either velocity [9-13, 18] or pressure boundary conditions [5, 25] for inlet boundary conditions. Both are described below for comparison purposes. In the present study, as described below, a time-varying velocity boundary condition was selected for the inlets, with each bronchial surface modeled as an inlet. The trachea was a stagnation pressure outlet, and all the other surfaces were modeled as rigid walls.

2.3.1 Pressure Boundary Conditions

A concept used in respiratory studies is known as airway resistance, which is defined as the ratio of driving pressure to flow rate. Different sections of the respiratory system have varying levels of resistance [6, 10, 18, 25]. In order to use a pressure boundary condition to induce an appropriate flow rate, knowledge of the total resistance of the respiratory system would be needed, as well as the portion of the overall resistance found in the area under consideration. Additional complications can arise due to resistances that change based on flow direction, pressure magnitude, or geometric scaling.
2.3.2 Velocity Boundary Conditions

Alternatively, other studies have used velocity boundary conditions, including uniform velocity profiles [9, 13], fully-developed profiles [10-12], as well as a profiles consistent with unsteady effects in a circular pipe [6]. The assumption of any profile shape is limited by how close a real profile would match up if a greater portion of the respiratory system were to be included, as the portions of the respiratory system upstream from the velocity inlet would affect the profile shape. In consideration of both commonly-used methods and their shortcomings and strengths, a velocity boundary condition was selected for the present studies with a uniform profile across the inlet.

2.3.3 Time-Varying Waveforms

Three breathing waveforms were explored. One breathing waveform, hereafter referred to as the biomimetic waveform, was based on the pressure waveform obtained from a zebra finch during quiet respiration. The other two breathing waveforms were based on periods of constant velocity for both inspiration and expiration, with a short period of transition between the constant velocity portions. These are described below.

Biomimetic Waveform

The biomimetic waveform was obtained from the signal reported in [31] that was acquired using a piezoresistive pressure transducer inserted into the left thoracic air sac of an adult male zebra finch at rest. A portion of the signal is shown in Fig. 2.3. No attempt had been made to calibrate the pressure values by the study authors, thus all units were left in Volts.

Six breathing pressure waveforms were then ensemble averaged, and a Fourier series approximating the resulting waveform was calculated. The first six terms of the series fit the
ensemble average well (see Fig. 2.3), and were used as a basis for the shape of the biomimetic waveform:

\[
V(t) = 0.075 \cos\left(\frac{2\pi t}{T} - 1.8546\right) + 0.007 \cos\left(\frac{4\pi t}{T} - 0.439\right) + \\
0.0196 \cos\left(\frac{6\pi t}{T} - 1.3507\right) + 0.0026 \cos\left(\frac{8\pi t}{T} - 0.1053\right) + \\
0.0055 \cos\left(\frac{10\pi t}{T} - 0.9668\right) + 0.0014 \cos\left(\frac{12\pi t}{T} + 0.0281\right)
\] (2.5)

where \( V(t) \) is the resulting amplitude of the signal \([V]\), \( t \) is the elapsed time \([s]\), and \( T \) is the period of the signal \([s]\). The pressure waveforms in every air sac in the lung-air-sac system of birds has been shown to be similar in shape to every other sac, and only differ from each other by less than one-tenth of the overall pressure magnitude [32]. A measurement in the thoracic sac therefore yielded a waveform of nearly equivalent shape and magnitude as compared to anywhere else in

Figure 2.3: Left: Pressure signal taken from the left thoracic air sac of a zebra finch from Franz and Goller (2001) [31]. Large amplitude oscillations, such as those seen between about 11 and 14 s, correspond to periods during which the zebra finch was singing. Smaller oscillations (amplitude of about 0.1 V) correspond to normal, resting breathing. The ratio of pressure magnitude between resting and singing (which here is about 10) is consistent with other literature, which reports ratios of up to 40 [30]. Only resting breathing was used for the ensemble average. Right: ensemble average along with a six-term Fourier series fit.
the lung-air-sac portion of the respiratory system. Although this waveform was derived from a pressure signal, not velocity, a boundary condition with a time-varying velocity profile using the shape of the biomimetic pressure waveform was used. This choice was justified by the quasi-steady nature of the signal, as discussed in Sec. 1.4.1 and Sec. 3.6. A study examining the pressure breathing waveform of a goose shows a similar shape [33]. While the exact shape of the waveform almost certainly does vary by species, and even within a species, the similar characteristics shown by both the zebra finch here (~10 g) and the goose (~5,000 g) [33] examined indicates that the shape of the waveform used here is likely closer to real breathing profiles than a sinusoidal approximation, the latter of which has been used in many unsteady studies of the respiratory system of humans and other animals [5, 6, 10, 11].

**Constant Velocity Waveforms**

In order to better understand the effect of breathing pattern waveforms on WSS, two different waveforms were tested in addition to the biomimetic waveform. These waveforms, shown in Fig. 2.4, were predominately constant velocity, with the constant velocity inspiration and constant velocity expiration portions separated by a transient portion that lasted one-tenth of the overall period. One waveform featured inspiration and expiration portions of equal duration. The other waveform had a longer expiration than inspiration by a ratio of three to one. These waveforms are hereafter referred to by their expiration to inspiration ratios (1/1 and 3/1, respectively). Real birds tend to have expiration periods that are of a longer duration than their inspiration periods [32, 33]. Testing constant velocity waveforms with different durations was intended to help quantify how both waveform shape and duration affected WSS.
2.4 Mesh

A depiction of the mesh for a 30 g bird with a 40° branching angle is shown in Fig. 2.5. All models featured a polyhedral mesh with numbers of cells ranging from 350,000 to 4,840,000, depending on what was needed for mesh convergence (further discussion in Sec. 2.6). All models also featured an extruded mesh off of the trachea and bronchi boundaries, extending the overall

![Figure 2.5: Depiction of a typical mesh. This mesh contained 744,759 cells. Red dashed lines indicate the location at which the pressure boundary condition (trachea) and velocity boundary condition (bronchi) were applied.](image)

![Figure 2.4: Breathing waveforms: biomimetic (left), 1/1 equal inspiration and expiration (center), and 3/1 sharp inspiration (right) waveforms. All were scaled to have matching tidal volumes. Waveforms were applied to the bronchial inlet boundaries (refer to Fig. 2.1). Inspiration portions correspond to negative velocity values.](image)
length of the trachea and bronchi slightly past the values given by the allometric equations (see Fig. 2.5). The extruded meshes at inlets and outlets helped to create a more representative computational domain by improving flow accuracy [Star CCM+ User Guide] and was found to help facilitate convergence. Ten layers of extruded mesh cells were used, which extended 5 mm past the geometries’ original boundaries. The pressure and velocity boundary conditions discussed in Sec. 2.3 were applied at the surfaces of the trachea and bronchi before the level of the extruded cells (see red lines in Fig. 2.5). The mesh near the TBJ, extending a distance of 0.25 times the tracheal length, was refined at half the base size in order to better capture flow patterns in the TBJ.

2.5 Solver Settings

Three-dimensional computational simulations of the avian airway during respiration were developed and conducted using the commercial CFD package STAR-CCM+ (Siemens). Peak Reynolds numbers (refer to Eq. 1.1) were calculated using the average velocity magnitude across the trachea during peak flow conditions, and using the tracheal diameter as the characteristic length. The fluid was air with density 1.184 kg/m³ and dynamic viscosity 1.855×10⁻⁵ Pa·s. The largest Reynolds number was 1340 for a simplified geometry simulation, which had a peak velocity of 1.25 m/s and a tracheal diameter of 1.68 cm. Because this value is below the expected turbulence threshold of 2,100 for internal flows [20], the flow was modeled as laminar. Peak flow rates, calculated using the peak average velocity of 1.25 m/s, resulted in a Mach number of 0.007, well below the commonly accepted transonic value of 0.3, so a constant density assumption was used. The flow was solved using a segregated, implicit, unsteady approach using first-order temporal discretization.
2.6 Verification

To verify simulations for both grid and time step independence, WSS at a point near the lateral side of the TBJ was monitored over the course of one full breathing cycle (see Fig. 2.6). Preliminary runs had identified the location chosen as a region that experienced high variations in WSS, indicating that it was a good location for verification studies. The simulation was considered to be grid- or time-independent when the maximum magnitude of the difference between the WSS values of successively refined mesh or time step sizes was less than 1% of the full scale range (defined as the difference between the maximum and minimum WSS values at any given time over one breathing cycle). This is shown in Eq. 2.6:

\[
\frac{|(WSS_2 - WSS_1)_{max\ difference}|}{WSS_{2,\max} - WSS_{2,\min}} < 0.01
\]  

(2.6)

where the subscript “2” denotes the more refined simulation, and “1” denotes the less refined simulation. WSS is calculated based on the gradient of velocity at the wall and is thus very sensitive to velocity field changes. It is therefore a good indicator of convergence. Additionally, because the primary output of interest in the simulations was WSS, it was logical to verify simulations based on this parameter. Results for all masses are shown in Fig. 2.7.

Figure 2.6: WSS was monitored over the course of one full breathing cycle at the location shown in red in order to verify grid and time convergence.
As seen in Fig. 2.7, the required grid count increased with increasing mass. Additionally, time step size was seen to have relatively little impact on convergence. Wo numbers for the simulations ranged from 0.34 (3 g) to 1.58 (30,000 g), with the 3,000 g bird being the first to exceed Wo = 1, the traditional limit for being able to use the quasi-steady approach (see Sec. 1.4.1.

Figure 2.7: Grid (left column) and time (right column) independence for the (starting from the top row) 3 g, 30 g, 300 g, 3,000 g, and 30,000 g bird mass simulations. Units for the legend entries are cell count (left column) and milliseconds (right column). Converged simulations are all shown as the dashed red line.
and Eq. 1.2). The time independence trends confirm this prediction, and the final step size was chosen based on the desired number of points per breathing cycle for the results rather than on a need based on time independence requirements.

2.7 Validation

The simulations were validated in a silicone airway model using particle image velocimetry (PIV) using a glycerol/water mixture seeded with titanium dioxide particles (https://pvsoap.com, Product ID: RM-TDR, CAS#: 13463-67-7). As the majority of the simulations were quasi-steady (Wo < 1), steady-state expiration and inspiration were used to validate the simulations. The PIV system consisted of a double-pulsed laser (New Wave Laser, Solo II-15 Nd:YAG, Fremont CA) and a synchronized camera (LaVision Imager Intense, Goettingen, Germany). A matched index of refraction method similar to that of Farley & Thomson [34] was followed.

2.7.1 Geometry

Figure 2.8 shows an image of the silicone airway model used for validation. A sacrificial model of the simplified airway (see Sec. 2.1) was first 3D printed (Prusa i3 MK2, Prusa Research SRO) out of PVA plastic. The model was then covered in a PVA-based synthetic glue (Elmer’s Washable Clear Glue, Newell Rubbermaid) and sanded in order to provide smooth edges. Clear, flexible PVC tubing with a 1 cm inner diameter was placed around the ends of the trachea and bronchi. Transparent silicone (QSil 216, QSi | Quantum Silicones) was mixed and vacuumed to remove air bubbles. The sacrificial model was suspended in an acrylic box, and the silicone was poured around the model. The silicone was allowed to cure. After curing, the PVC tubing was removed and the sacrificial portion was dissolved using water, leaving only the airway within the
The airway had a tracheal diameter of 1 cm and a tracheal length of 4.5 cm. The bronchi had diameters of 1 cm, were 4.5 cm long, and had a branching angle of 80°. The pessulus had a radius of curvature of 1 cm. Transparent and flexible PVC tubing with an inner diameter of 1 cm was glued onto the ends of the models with silicone adhesive (Sil-Poxy, Smooth-On, Inc.) so as to have a smooth transition from silicone model to tubing, and consequently minimize flow disturbances.

### 2.7.2 Index of Refraction Matching

Prior to each test, the glycerol (Fisher Chemical) and water mixture ratio was tuned to match index of refraction of the silicone in order to enable optical access without distortion due to index of refraction mismatch. An initially glycerol-heavy mixture was slowly diluted with water until the index of refraction of the mixture matched that of the silicone. A grid was placed behind the model to assess how well the index of refraction matched; see Fig. 2.9. The index of refraction was considered to be matched when the lines on the grid were not distorted behind the boundaries.

Figure 2.8: Picture of the silicone model containing the airway with a grid placed behind to aid in index of refraction matching.
between the fluid and the model. While the exact ratio varied slightly between tests, an approximate volumetric ratio of 53/47 of glycerol to water was used. The dynamic viscosity and density of the mixed fluid were calculated according to Cheng (2008), and found to be $6.38 \times 10^{-3}$ Pa·s and 1133 kg/m$^3$, respectively.

### 2.7.3 Experimental Setup

Figure 2.10 illustrates the experimental setup in which the silicone model was part of a closed loop with a pump driving the flow. After matching the index of refraction as described in Sec. 2.7.2, a sheet of black paper was placed behind the silicone for maximum particle visibility without background interference. The camera was then focused on the model itself. The laser sheet was positioned along the coronal plane that bisected the model. Titanium dioxide seeding particles were added to the reservoir until the desired seeding density was achieved. The lens and focus were adjusted. At this point, the pump was turned off, and the valve closed.

For data acquisition, flow through the silicone model was driven by a linear actuator attached to a 30 mL syringe. The fluid was driven at a constant flow rate of 1.90 mL/s. In order to remove any significant transient effects, imaging did not begin until the actuator had been moving for one
second. Forty image pairs were captured at a rate of 4.95 Hz, with a 10 ms delay between the first and second image in each pair. A sample image is shown in Fig. 2.11. The process was repeated four additional times for each flow direction, for a total of 200 image pairs in each flow direction. Image resolution was 1376×1024 pixels. Following data collection, the system was drained and the silicone model was cleaned in order to prevent seeding particle deposition on the airway walls of the model. Velocity fields were analyzed using an FFT-based cross-correlation algorithm (DaVis V.7). A multi-pass scheme was implemented, which initially used a 128×128 pixel interrogation window with 50% overlap, and concluded with a 32×32 pixel interrogation window size with 75% overlap. The velocity fields for all image pairs were then averaged.

Figure 2.10: Schematic of the PIV setup. The valve connecting the pump was turned off prior to testing such that water movement was controlled exclusively by the linear actuator.
2.7.4 Dynamic Similarity

Reynolds numbers of the expiration and inspiration PIV measurements matched those of a set of steady simulations designed for validation purposes. The steady inspiration and expiration simulations were set up using the 30 g simulation size. With diameters, densities, and viscosities set, velocities were chosen to match Reynolds numbers between simulations. The fluid expelled by the linear actuator had a flow rate of 1.90 mL/s, which resulted in a Reynolds number of 43. This resulted in an average air velocity of 0.411 m/s in the trachea of the CFD model, and an average glycerol/water mixture velocity of 0.0242 m/s in the trachea of the experimental model.

2.7.5 Validation Results

Inspiration

The inspiration PIV and CFD results, shown in Fig. 2.12, match well in several qualitative respects. The jet of fluid impinges on the pessulus, with the area just above the pessulus being a
stagnation point. An additional region of low velocity is located near the lateral walls of the TBJ, just below where the bronchi first start to branch out. Velocity magnitudes (scaled by Reynolds number) in the computational model, while generally slightly be low the magnitudes seen in the experimental results, are visually similar in magnitude.

A quantitative analysis of the velocity magnitudes confirms the similarities seen in Fig. 2.12. Velocities were extracted along the line marked “A-A” in the right bronchus, as shown in Fig. 2.13. For the computational model, this was achieved using a line probe to directly extract the velocities. For the experimental model, this was achieved using a grayscale figure and examining the intensity values along the same line (refer to Appendix A). Results are shown in the right section of Fig. 2.13. Velocities were scaled relative to the largest intensity for the experimental model and relative to the largest velocity for the computational model. Values differed by an
average of 2.1%, with the maximum difference calculated as 4.6% when compared to the maximum normalized velocity. The magnitude of the normalized velocity gradient of the CFD results was about 17% higher at the lateral wall, and about 11% lower at the medial wall (see Table 2.1, and Appendix A for the calculation thereof).

Expiration

The PIV and CFD results for expiration were qualitatively similar, with a triangular region of lower velocity above the pessulus (see Fig. 2.14), and greatest velocity magnitudes just above the TBJ. The velocity profiles for the expiration experimental results were extracted in the same manner as the inspiratory results. The expiration results were on average quantitatively even closer than the inspiration results (see Fig. 2.15), with results differing by on average 1.3%, with a maximum difference of 6.8%, which occurred at the lateral wall. The magnitude of the normalized velocity gradient of the expiratory CFD results was less than 1% lower at the medial wall, and about 3% higher at the lateral wall.

Figure 2.13: Grayscale results from inspiratory PIV used to calculate the velocity profile along the line marked A-A. (left). The graph on the right shows the agreement between PIV results and the CFD results. $V_{\text{max}}$ was 0.027 m/s for the PIV results and 0.46 m/s for the CFD results.
Table 2.1: Comparison of gradients at the wall during inspiration and expiration for CFD and PIV

<table>
<thead>
<tr>
<th></th>
<th>dV/dr (1/s)</th>
<th>dV'/dr' (Non-Dimensional)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>PIV</td>
<td>CFD</td>
</tr>
<tr>
<td>Inspiration</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Medial</td>
<td>0.00134</td>
<td>0.450</td>
</tr>
<tr>
<td>Lateral</td>
<td>0.00075</td>
<td>0.191</td>
</tr>
<tr>
<td>Expiration</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Medial</td>
<td>0.00122</td>
<td>0.351</td>
</tr>
<tr>
<td>Lateral</td>
<td>0.00118</td>
<td>0.353</td>
</tr>
</tbody>
</table>

Figure 2.14: Expiratory PIV results (left) and CFD results (right). Local Reynolds numbers were computed by using the velocity at any given point rather than the average velocity magnitude.

2.8 Conclusion

This chapter discussed settings and methods used for the simplified model simulations. Geometries, boundary conditions, and simulation parameters were described. Verification studies to establish grid and time independence were presented. Grid requirements increased with
increasing mass, while time step requirements was found to only be significant for the larger mass simulations, which was consistent with Womersley number predictions. PIV results were compared to CFD results for validation. A comparison of extracted velocity profiles taken across the diameter of the right bronchus differed the most near the walls, at which point PIV results predicted non-zero velocity. This implied that the difference seen near the wall was more likely an error in PIV measurements than an error in the CFD results. The results overall for the extracted profiles differed by less than 3% on average when compared to the maximum normalized velocities which indicated that the PIV results adequately validated the model.

Figure 2.15: Grayscale results from expiratory PIV used to calculate the velocity profile along the line marked A-A (left). The graph on the right shows the agreement between PIV results and the CFD results. $V_{\text{max}}$ was 0.024 m/s for the PIV results and 0.40 m/s for the CFD results.
3 WALL SHEAR STRESS AND VELOCITY PATTERNS IN SIMPLIFIED GEOMETRY MODELS OF THE AVIAN AIRWAY

The focus of this chapter is on the results obtained from CFD simulations using the simplified geometry. Simulations included variation in the size of the respiratory system using allometric equations, breathing patterns, and bronchial branching angles, as described in Chapter 2. Each of these parameters were varied parametrically, changing only one parameter at a time. All parameters were found to impact WSS as discussed below.

3.1 Measurement Metrics for Comparison

Several metrics were gathered in order to assess the effects of different parameters on WSS and velocity patterns. Velocity was examined along the coronal plane that bisected the model, as well as in a plane perpendicular to one bronchus just inferior to the curve of the pessulus (see ‘A’ in Fig. 3.1). WSS was evaluated within the TBJ, as well as along half of the tracheal and bronchial length (Region ‘B’ in Fig. 3.1). Both average and maximum WSS were evaluated over Region B for each time step. The approximate locations of maximum and minimum WSS during inspiration and expiration phases were also studied and are noted in selected sections below.

3.2 Typical Results

The simulation parameters of the “general case” from which parameter variations were implemented were: 30 g mass, 40° branching angle, and biomimetic breathing profile. Contours
Figure 3.1: Depiction of the model where velocity was evaluated perpendicular to the bronchi just inferior to the pessulus (A), and the region over which maximum and average WSS were evaluated (B). For the purposes of this study, the TBJ is defined as the region ranging from where the trachea first grows past its nominal diameter extending to the end of the radius of curvature of the pessulus.

of velocity and WSS over a typical breathing cycle for the general case are shown in Fig. 3.2. While all models did exhibit variability, some commonalities existed across all simplified geometry simulations. All simulations exhibited maximum WSS values near the lateral side of the TBJ during expiration (see Inset A in Fig. 3.2), and maximum values near the pessulus during inspiration (Inset B), with the highest overall values being those near the pessulus during inspiration. Maximum WSS values were typically between three to ten times greater than the average WSS values during peak inspiratory flow conditions. The location of the minimum WSS across all simplified geometry simulations during expiration was the center of the pessulus (Inset A), while the location of the minimum WSS during inspiration varied, but was always either along the lateral wall of the TBJ or on the lateral side of the bronchi (Inset B). Secondary flow patterns,
defined here as flow perpendicular to the axis of either the trachea or the bronchi, were small \((V_{\text{axial}} \gg V_{\text{perpendicular}})\) across all simulations, with the exception of during inspiration in the bronchi. During inspiration in the bronchi, the secondary flow traveled laterally; however, the exact patterns with which the flow behaved varied, and are further discussed in subsequent sections.

Figure 3.2: Top row: Close-up of the WSS pattern during expiration (Inset A), the breathing profile with time/period labels (center), and a close-up of the WSS pattern during inspiration (Inset B). In the insets, the red dashed lines refer to the region of minimum WSS, while the orange dashed lines show regions of maximum WSS. Middle row: Velocity profiles for the general case (30 g mass, 40° branching angle, biomimetic breathing pattern) for six different times of the breathing profile; i.e., 1/6 refers to one-sixth of a period. Bottom row: WSS profiles at the same time instances as the middle row.
3.3 Effect of Mass on Wall Shear Stress and Velocity Patterns

The influence of mass on average WSS can be seen in the top-left graph of Fig. 3.3. The average WSS value was roughly symmetric between expiration and inspiration. This is because

![Graph showing average and maximum WSS for each bird size during one breathing cycle.]

![Velocity profiles for birds of different sizes at peak inspiratory flow (dotted line in the graphs). The red lines illustrate where secondary profiles were evaluated.]

![WSS profiles evaluated for each mass at the same time as the middle row.]

Figure 3.3: Top row: Average and maximum WSS for each bird size during one breathing cycle. Expiration precedes inspiration. Middle row: Velocity profiles for birds of different sizes at during peak inspiratory flow (dotted line in the graphs). The red lines illustrate where secondary profiles were evaluated. Bottom row: WSS profiles evaluated for each mass at the same time as the middle row.
the velocities based on the breathing profiles (refer to Fig. 2.4) were roughly of the same magnitude and roughly symmetric between expiration and inspiration. Consequently, the average WSS values were dominated by sections of the model with relatively moderate velocity gradients (i.e., regions away from the TBJ) and generally similar velocity profiles, albeit in opposite directions (see similarities in velocity and WSS patterns in regions away from the TBJ in Fig. 3.2 between times 1/6 and 4/6, between 2/6 and 5/6, and between 3/6 and 6/6). Larger sizes exhibited lower average WSS, with WSS generally dropping by nearly half for each successively larger simulation (see the bottom row as well as left graph of Fig. 3.3). Although larger sizes yielded slightly faster average velocities, with each mass size having an average velocity magnitude increase by about 16% at any given point in the breathing period (see the middle row of Fig. 3.3 and Eqns. 2.1, 2.2, and 2.4), the cross-sectional area grew faster than the average velocity, resulting in lower WSS values on average.

On the other hand, the maximum WSS profiles varied greatly in magnitude from expiration to inspiration (see the graph on the top right of Fig. 3.3). During expiration, the region of maximum WSS was along the lateral edge of the TBJ, near where the bronchi first begin to branch out, as described in Sec. 3.2. On the other hand, during inspiration, the jet of air from the trachea impinged on the pessulus, creating sharper velocity gradients and consequently a larger magnitude of WSS than during expiration. Between the different sizes, the jet of air entering the TBJ during inspiration generally impinged more strongly on the pessulus during inspiration for larger sizes (as seen by the faster velocities with increasing mass in Fig. 3.3), resulting in higher maximum WSS values. This continued to be the case until the 30 kg size, in which the volume of the TBJ grew large enough that the jet no longer impinged quite as strongly in a relative sense, noting that diameters approximately doubled in size with each increase in mass, while velocity only grew by about 16%.
Additionally, as the mass increased and the geometry consequently increased in size, a region of recirculation formed along the lateral sides of the TBJ (see Fig. 3.4). The velocity within each bronchus was increasingly asymmetric with increasing mass, with higher velocities near the medial surface of the bronchi (middle row of Fig. 3.3). This in turn caused higher medial surface WSS values in the larger simulations.

The location of minimum WSS during inspiration was increasingly in the inferior direction with increasing mass but was always on the lateral side of the TBJ or the bronchi. For simulations that were of 300 g and larger, the location coincided with where the region of reversed flow ended (see Fig. 3.4), which formed a pattern like a saddle point. The two smallest sizes (3 g and 30 g) did not feature any reversed flow during peak inspiration, and the minimum WSS location was near the top of the TBJ.

![Velocity profiles with vectors indicating flow direction for simulations of different masses. Red dots indicate position of lowest WSS, as well as the boundary of where reversed flow occurred for simulations larger than 30 g. Actual vector density was approximately four or more times greater than shown here (refer to cell count in Fig. 2.7).](image-url)
Secondary flow patterns during inspiration in the bronchi were also affected by mass (see Fig. 3.5). The smallest simulation (3 g) did not have secondary vortices visible in the bronchi. However, vortices were present in the simulations of masses from 30 g and larger. As size increased, the strength of the flow in the medial direction increased in magnitude. While the magnitude of the flow in the medial-lateral direction also increased, the flow became more circumferential rather than purely lateral in direction.

3.4 Effect of Branching Angle on Wall Shear Stress and Velocity Patterns

The influence of bronchial branching angle on average and maximum WSS can be seen in Fig. 3.6. Because mass and breathing patterns remained constant in the set of simulations in which the branching angle was varied, the average WSS was very similar for all branching angles (see the
top left graph on Fig. 3.6), as the average WSS was dominated by the instantaneous velocity and tracheal and bronchial diameters. Average WSS values differed by less than 10% when compared to the maximum average WSS. During expiration, branching angle had little effect on maximum WSS (see the top right graph on Fig. 3.6, from 0 to approximately 0.55 seconds).

Figure 3.6: Top left: Average WSS and Top right: maximum WSS for each angle-varying simulation. Expiration preceded inspiration. Middle row: Velocity and Bottom row: WSS profiles during peak inspiratory flow (as indicated by the dotted lines in the graphs). The red line in each velocity profile shows where the velocity fields in Fig. 3.7 were evaluated.
While maximum WSS was not found to be strongly affected by branching angle during expiration, the branching angle was found to be important in determining the maximum WSS during inspiration. Two factors were found to be important regarding the magnitude of maximum WSS during inspiration: the vertical size of the TBJ (defined here as the point from which the trachea begins to expand from its nominal diameter extending to the pessulus), and the sharpness of the pessulus (defined as the arc length defining the pessulus, as the pessuli of the branching angle simulations had the same radius of curvature). A sharp pessulus (e.g., 40°) did not disturb the flow to the same extent as a dull pessulus (e.g., 90°), which allowed for the jet of air entering the bronchi from the trachea to be relatively undisturbed in direction, allowing for steeper velocity gradients close to the pessulus. A large TBJ (e.g., 15°) allowed the incoming jet to spread and consequently approach the pessulus at a lower velocity, resulting in lower WSS at the pessulus. The 15° simulation had the sharpest pessulus, but also had by far the largest TBJ, and consequently the lowest maximum WSS value. The middle angles (40° and 65°) had a comparable TBJ depth as the 90° case, but a sharper pessulus than the 90° simulation.

No reversed flow was seen during inspiration in any angle-varying simulations. The location of minimum WSS moved in the inferior direction with decreasing angle, but remained within the TBJ for all angles (see Fig. 3.6, bottom row). Considerable variation was seen in the secondary flow profiles just inferior to the pessulus during inspiration (Fig. 3.7). The smaller the angle, the smaller the magnitude the flow was in the medial direction. This was in part due to the size of the TBJ across the angle-varying simulations. Similar to what was discussed in the first paragraph of Sec. 3.4, the lower angle simulations had a larger TBJ, and consequently, the flow had longer to turn and become more aligned in the axial direction of the bronchi. The magnitude of the in-plane velocity was therefore of a lower magnitude than the larger-angled simulations.
3.5 Effect of Breathing Pattern on Wall Shear Stress and Velocity Patterns

The influence of breathing pattern waveform on average and maximum WSS can be seen in Fig. 3.8. Each of the three simulations (the biomimetic, the 1/1 ratio, and the 1/3 ratio) exhibited a similar average WSS profile shape as its corresponding breathing profile (see Fig. 2.5). For the 1/1 ratio simulation, average WSS values were slightly higher during expiration than the average WSS magnitude during inspiration (see the left graph of Fig. 3.8) despite having equal average tracheal velocities (see Fig. 2.5). The average WSS for the 3/1 ratio simulation during inspiration was more than three times higher than the expiratory average WSS, despite the velocity magnitudes being only three times as large, indicating a non-linear WSS to velocity magnitude relationship. The
resulting average WSS for the 1/1 ratio and 3/1 ratio waveforms exhibited small peaks at the end of their transient portions, before settling into a steady state. This trend is further discussed in the next section.

With maximum WSS results, shown on the right of Fig. 3.8, expiration maximum values are much lower than inspiration, which is in line with the typical results across all simulations (Sec. 3.2). Of note is that the highest maximum WSS seen across all the simplified geometry simulations was that of the 3/1 ratio simulation, which reached a magnitude of 0.70 Pa. The principle cause for the high WSS value is likely due to the elevated inspiratory velocity of the 3/1 simulation as compared to the other simulations (see Fig. 2.4).

The secondary vortices seen in most other simulations perpendicular to the flow (see Fig. 3.5 and Fig. 3.7) were present during peak inspiratory flow of the 3/1 ratio simulation and the biomimetic simulation, but not the 1/1 ratio simulation (see Fig. 3.9). This is consistent with trends seen in other simulations (vortices were not seen in the 3 g, which had a lower peak Reynolds number, but were seen in larger masses), and reveals that the vortices seen are dependent on Re.

Figure 3.8: Average WSS (left) and maximum WSS (right) during one breathing cycle. Inspiration preceded expiration.
number and not Wo number, as the vortices were observed during the steady-flow inspiratory portion of the 3/1 ratio waveform. The minimum WSS location was unchanged for all breathing pattern simulations (see Fig. 3.9).

3.6 Unsteady Effects

As discussed in Section 1.4.1, the Womersley number relates viscous effects to the frequency at which the fluid is oscillating. As discussed below, the impact of unsteadiness is quantified in both the breathing pattern simulation set (see Sec. 3.5), which had transient and steady waveform sections, as well as in the mass-varying simulation set (see Sec. 3.3) which featured Womersley numbers from 0.34 to 1.58.

Figure 3.9: Top row: Red dot indicates minimum WSS location for each breathing pattern during peak inspiration flow rate (at 0.38 s for the biomimetic waveform simulation, and at 0.2 seconds for the 3/1 and 1/1 ratio simulations; see Fig. 2.4). Bottom row: Velocity cross sections for each breathing pattern simulation at the location marked for the 30 g simulation in Fig. 3.3.
3.6.1 Unsteady Effects in Breathing Pattern Simulations

The effect of unsteadiness on WSS and velocity can be most clearly seen in the simulations that featured constant velocity waveform sections (see Fig. 2.4). While the waveforms were predominately constant velocity, there were transient portions of the waveforms when switching from inspiration to expiration and from expiration to inspiration, which each lasted 1/10 of the full period (refer to Sec. 2.3.3 and Fig. 2.5). The Womersley number requires the frequency for a full cyclic cycle (i.e., starting and ending at the same velocity condition) so the combined time for both transient sections (as if the transient sections together were the complete cycle) combined for 1/5 of the full period. Using this frequency to calculate the Womersley number changes the Womersley number from 0.50 for the whole period to 1.12 when calculated for the transient sections of the period only (refer to Eq. 1.2). As Womersley number increases, the velocity profile is expected to become more plug-like and less parabolic. By comparing the velocity profile at the end of one of the transient portions of the breathing waveform with the velocity profile after the flow has reached steady-state, the effect of unsteadiness can be better understood and quantified. To do so, a velocity profile was taken across the trachea halfway between the trachea outlet and the TBJ, along the coronal plane that bisected the model, at the end of the transient inspiration to expiration period of the 3/1 ratio waveform model (see Fig. 3.10b, blue circle) and again once the WSS had reached steady state (see Fig. 3.10b, red circle). The results can be seen in Figure 3.10a.

While both profiles were mostly parabolic, the profile taken at the end of the transition section exhibited a slightly lower peak velocity and a slightly steeper slope near the wall when compared to a steady-state parabolic profile (Fig 3.10a) and hence higher WSS (Fig. 3.10b). The same process occurred for flow during the beginning of the inspiration constant velocity period: average
WSS peaked before settling into steady state. Refer to Table 3.1 for average WSS magnitude comparisons at the end of transition and at steady state for both the 1/1 ratio and 3/1 ratio simulations.

For maximum WSS, when switching from inspiration to expiration, the location of maximum WSS moved from the pessulus to the lateral sides of the TBJ, as described in Sec. 3.2. The lateral sides of the TBJ had a higher wall shear stress due to Womersley number effects (a more plug-like profile), and then settled into the steady-state values. While not as pronounced as the average WSS peaks, small peaks can be seen in maximum WSS graph (Fig. 3.8, right graph, at 0.275 seconds).

### Table 3.1: Percentage above or below the steady-state value at the end of the transient portions of the breathing waveform for the 3/1 ratio and 1/1 ratio simulations.

<table>
<thead>
<tr>
<th></th>
<th>Inspiration to Expiration</th>
<th>Expiration to Inspiration</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Average WSS</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1/1 Ratio</td>
<td>3%</td>
<td>3%</td>
</tr>
<tr>
<td>3/1 Ratio</td>
<td>17%</td>
<td>3%</td>
</tr>
<tr>
<td><strong>Maximum WSS</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1/1 Ratio</td>
<td>1%</td>
<td>-9%</td>
</tr>
<tr>
<td>3/1 Ratio</td>
<td>10%</td>
<td>-10%</td>
</tr>
</tbody>
</table>

Figure 3.10: Left: velocity profile in the trachea at the end of the transition and at steady-state between and after the inspiration to expiration switch for the 3/1 ratio expiration to inspiration simulation. Right: average WSS for the same simulation showing the time instances when the profiles were extracted. See Fig. 3.8 for the full average WSS profile.
for the 3/1 ratio simulation and at 0.59 seconds for the 1/1 ratio simulation). On the other hand, when moving from expiration to inspiration, the flow near the center of the profile, and consequently near pessulus, was slightly slower due to the Womersley number effects, and thus the WSS value increased as the flow became steady. These types of elevated WSS levels due to unsteady effects have also been observed in human respiratory studies that used square waveforms [9, 12]. Refer to Tab. 3.1 for maximum WSS magnitude comparisons at the end of transition and at steady state for both the 1/1 ratio and 3/1 ratio simulations.

3.6.2 Unsteady Effect in Mass-Varying Simulations

Womersley numbers for the range of masses tested ranged from 0.34 to 1.58, which is in line with the Womersley numbers expected for general animal respiration [19]. In order to quantify part of the effect of unsteady effects on WSS, the results of the mass-varying simulation set were compared to flow in a straight pipe. The flow in a straight pipe comparison assumed laminar, incompressible, fully-developed flow. The relative lengths of the trachea and bronchi, as well as the fact that the flow rates through the bronchi were only half as fast was taken into account (as the areas of the bronchi combined were twice that of the tracheal area). The equation for WSS under these conditions is given by [35]:

\[
\tau_w = \frac{8\rho V^2}{\text{Re}}
\]  

(3.1)

where \(\tau_w\) is the WSS [Pa], \(\rho\) is the density [kg/m\(^3\)], \(V\) is the average velocity [m/s], and \(\text{Re}\) is the Reynolds number. The results of comparing the average WSS during peak inspiratory flow with what laminar flow in a straight pipe would predict (as given in Eq. 3.1) with the same flow rate are shown in Fig. 3.11.
As can be seen, the results for sizes from 3 to 300 g (Womersley numbers from 0.34 to 0.735) match very well with Eq. 3.1, while Eq. 3.1 under predicts WSS in sizes from 3,000 to 30,000 g (Womersley numbers from 0.735 to 1.58), with increasing mass leading to increasing deviation. This can also be seen when comparing the actual average magnitude over the course of one breathing cycle to what Eq. 3.1 predicts for straight pipe flow (with the same assumptions as previously stated) for both the smallest size and the largest size, 3 g and 30,000 g, respectively (see Fig. 3.12). The 3 g average WSS profile matched straight pipe flow theory much more closely than that of the 30,000 g case. An extremely simplified theory could be used to predict average WSS for small masses, and thus saving relatively expensive computational time, but not in larger masses, primarily due to unsteady effects for the geometric configurations as presented in Sec. 2.1.

3.7 Conclusion

Bird mass, bronchial branching angle, and breathing pattern waveform were parametrically varied to assess their impact on velocity patterns and resulting WSS magnitudes. The pessulus was
identified as a region in which maximum WSS occurred during inspiration, and minimum WSS occurred during expiration. The lateral sides at the transition between the trachea and bronchi experienced the highest WSS during expiration. The location of the minimum WSS during inspiration varied with mass and branching angle. Unsteady effects were seen and quantified in both the mass-varying simulations, as well the breathing pattern simulations. WSS values were higher in magnitude than steady-state portions of simulations, as seen in the 3/1 and 1/1 ratio simulations. Mass generally increased the maximum WSS value experienced over the entire cycle, with the 3,000 g size experiencing the highest WSS value over the mass-varying simulations. Outlying angles (15° and 90°) experienced the lowest maximum WSS, with the 40° simulation experiencing the highest value over the angle-varying simulations. The breathing pattern simulations demonstrated that WSS values are closely linked with the instantaneous average velocity value, with the high velocity of the 3/1 ratio simulation having the highest WSS value seen both among the breathing pattern simulations as well as the highest seen in all the simplified

Figure 3.12: WSS results from the 3 g simulation (left) and 30,000 g simulation (right) compared to laminar flow theory (Eq. 3.1) using the biomimetic velocity profile velocity.
geometry simulations. Future work with the simplified geometries could account for the presence of vocal fold structures near the TBJ, which would constrict the airway and have a compliant oscillating region mimicking phonation.

The presence of relatively high and low WSS near the region of the TBJ in which the syrinx forms in extant birds (see Sec. 1.1) is not enough of itself to determine whether WSS contributed to the development of the syrinx, or what advantages a syringeal vocal fold position offers. However, if WSS were determined to have contributed to the development and advantages of a syringeal vocal fold position, the WSS values and trends herein presented form a basis from which to further study both the development and advantages of a syringeal vocal fold position from a fluid dynamics perspective.
4 WALL SHEAR STRESS AND VELOCITY PATTERNS IN SCANNED MODELS OF THE AVIAN AIRWAY

As discussed in Chapter 1, considerable variation is seen in the morphology of the avian syrinx [1]. Even among humans, variations in respiratory system morphology has been shown to affect the flow patterns during breathing [22-24]. While it would not be possible to fully sample the estimated 18,000 [29] species of birds, both for lack of available data as well as practical constraints of time and computing resources, comparing the results obtained from using the techniques presented in Chapters 2 and 3 with results from detailed models can serve to highlight similarities and differences that arise between using simplified versus scanned geometries. To that end, this chapter explores results from simulations using CT-scanned geometries and compares them to results obtained using the simplified geometry simulations described in Chapters 2 and 3.

4.1 Bird Selection

The respiratory system of a Buff-tailed Sicklebill hummingbird (Eutoxeres condamini) and a Little Swift (Apus affinis) were used for the scanned geometries. The hummingbird weighed 14.28 g, while the swift weighed 36.71 g. These species were chosen based on availability of scans of the tracheobronchial region that were conducted in situ, so as to best preserve the natural structure of the respiratory tract.
4.2 CT scan

All scans were performed by partner researchers at the University of Texas. Both birds were scanned in situ (dead but intact) and were arranged in a natural body position. The hummingbird scan was a detailed scan near the syrinx, which had isotropic voxels with 10 μm resolution. A sample of one image used is shown in Fig. 4.1. The little swift was scanned with a full-body scan, which had isotropic voxels with 44.8 μm resolution.

4.3 Geometry Extraction

The hummingbird scan was imported into ITK-SNAP, an open source medical image segmenting software program. In order to segment the region of interest, the image contrast settings were adjusted to clearly delineate the boundaries of the airway. An automatic segmentation algorithm was then used to segment the airway using a thresholding function. This region was imported into Autodesk Meshmixer (Autodesk, Inc.), which was then employed to make clean cuts approximately perpendicular to the trachea and the bronchi. The resulting surface

Figure 4.1: Image from hummingbird CT scan, provided by researchers at the University of Texas. The region in yellow depicts the area that was segmented. The red bar corresponds to a distance of 1 mm.
mesh was then imported into Star-CCM+. Depictions of the raw segmented geometry, as well as the cut geometry, are shown in Fig. 4.2. The overall tracheal and bronchial lengths that could be segmented were limited based on the available RAM. The little swift geometry was provided in a segmented format by partner researchers at the University of Texas who segmented the little swift airway using Avizo 9 (ThermoFisher Scientific). The geometry was cut using the same process described above for the hummingbird geometry. Depictions of the little swift airway after segmentation as well as after cutting are shown in Fig. 4.2.

### 4.4 Simplified Models and Allometric Equations

Simplified models (see Chapters 2 and 3) were generated using the allometric equations (Eqns. 2.3-2.4) with the same mass as the birds that were scanned (see Sec. 4.1). A comparison of geometric lengths and areas for simplified and scanned models is given in Table 4.1. As can be

![Figure 4.2: Hummingbird geometry extracted from ITK-SNAP (far left), and after cuts were made perpendicular to the trachea and bronchi (left center). Geometry provided for the little swift (right center) and after cuts were made perpendicular to the trachea and bronchi (right). The diameter of the trachea of the hummingbird was about 1.15 mm, and the diameter of the trachea of the little swift was about 1.29 mm.](image-url)
seen from Table 4.1, the allometric equations overpredicted the tracheal dimensions of both birds. Allometric equations for scaling the frequency and tidal volume (Eqns. 2.1 and 2.2) were used to generate the breathing patterns, as described in Chapter 2. The biomimetic profile was used for all simulations. Bronchial branching angle was made to match as closely as possible the scanned geometries. The little swift had a branching angle that was simple to measure and was measured at approximately 60°. The branching angle of the hummingbird, however, was more difficult to ascertain. On one hand, the bronchi initially separated at a wide angle, approximately 90°; however, the bronchi soon curved back around, and were nearly parallel (see Fig. 4.2). Therefore, two simplified simulations were tested for the hummingbird geometry, one with a 15° and one with a 90° branching angle.

4.5 Measurement Metrics for Comparison

The metrics gathered and compared in the scanned geometries were identical to those gathered for the simplified geometries (e.g., maximum WSS, average WSS, location of maximum
and minimum WSS, and velocity patterns; see Sec. 3.1) with a few necessary modifications. Unlike the simplified models, the scanned geometries were not symmetric across any plane. Efforts were made to derive a plane that closely bisected the airways into posterior and inferior halves despite the asymmetry. The hummingbird geometry, which was noticeably curved, was bisected using a cylindrical surface with a 10 mm radius of curvature, while the little swift was able to be bisected using a plane (see Fig. 4.3). These sections were used in order to evaluate velocity profiles. Additionally, unlike the simplified geometry, several regions of low WSS of a similar magnitude were found to exist rather than a clear single location as was found in the simplified geometry simulations (see Sec. 3.2), which is discussed in further detail below.

Figure 4.3: Cross sections along which velocity contours were explored for the hummingbird (left) and little swift (right) simulations.
4.6 Other Settings

In lieu of using velocity boundary conditions on each of the bronchi, a mass flow boundary condition was used at the superior end of the trachea. As the scanned models were not perfectly symmetric, forcing each bronchus to have identical flow rates would not be true to life, as flow in non-identical geometries will not be identical due to differing resistances [18, 35]. All other computational settings not discussed in Chapter 4 (e.g. solver settings, mesh settings, etc.) were identical to those discussed in Chapter 2.

4.7 Verification

The scanned hummingbird and little swift geometries were verified for both time and grid independence, in the manner described in Sec. 2.6. WSS was monitored at a point near the TBJ over one full breathing cycle (see Fig. 4.4 for the precise positioning of the points at which WSS was monitored in both hummingbird and little swift geometries). The scanned simulations were considered converged after the maximum WSS difference at any point in the breathing cycle was

Figure 4.4: Location indicated by a red dot is where the WSS value was monitored over one breathing cycle for the hummingbird (left) and little swift (right).
less than 1% of the full scale range (defined as the difference between the maximum and minimum WSS values in the cycle). Results for the scanned hummingbird and little swift simulations can be seen in Fig. 4.5.

Rather than perform additional grid- and time- independence studies on the simplified geometries (further than those undertaken in Chapter 2), grid- and time- independence for the simplified geometry simulations was conservatively taken as the same number of cells and time step size needed for the geometry just larger, as determined in Chapter 2. The 14 g simplified hummingbird simulation used the same number of cells and time step size as what had been found

Figure 4.5: Grid (left column) and time (right column) independence for the little swift simulation (top row) and hummingbird simulation (bottom row). In each graph, the dashed red line represents the converged solution. Entries in the legend of the grid convergence column (left) are in units of cells, while entries in the time convergence column (right) are in units of milliseconds.
to be needed for the 30 g simulation, and the 37 g simplified little swift simulation used the same number of cells and time step that had been found to be needed for the 300 g simplified simulation (see Fig. 2.7 for all time and grid independence values for the 30 g and 300 g simulations). The cell count required for grid converged solutions was much higher than that needed for the simplified geometries (see Sec. 2.6).

The trend seen with the simplified models of larger bird sizes requiring a greater number of cells (see Fig. 2.7 and Sec. 2.6) was observed with the scanned simulations as well (see Table 4.2 and Fig. 4.5), with the larger little swift geometry requiring twice as many cells to reach convergence. The relative insensitivity of the time step size to reach convergence was likewise seen here, which is consistent with the Womersley numbers, calculated for the hummingbird and little swift, respectively, as 0.44 and 0.52, predicting that a quasi-steady assumption could be used (refer to Eq. 1.2 for Womersley number calculation). A summary of the grid count and time step used is presented in Table 4.2.

### 4.8 Scanned Hummingbird vs. Simplified Simulations

A depiction of WSS and velocity profiles during a breathing cycle in the scanned hummingbird geometry is shown in Fig. 4.6. The scanned hummingbird simulation had high WSS throughout in the region of the hummingbird’s vocal folds (see Inset B in Fig. 4.6). Excluding this region, high

<table>
<thead>
<tr>
<th></th>
<th>Hummingbird, Scanned</th>
<th>Hummingbird, Simplified</th>
<th>Little Swift, Scanned</th>
<th>Little Swift, Simplified</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cell Count x 1000</td>
<td>1,500</td>
<td>750</td>
<td>3,000</td>
<td>3,000</td>
</tr>
<tr>
<td>Time Step (ms)</td>
<td>3.6</td>
<td>0.5</td>
<td>7.7</td>
<td>10</td>
</tr>
</tbody>
</table>
WSS is notably higher near the pessulus during inspiration due to the incoming air impinging on the pessulus (see velocity and WSS profiles at times 4/6 through 6/6 in Fig. 4.6), similar to what is seen in the simplified geometries (see Fig. 3.2, velocity and WSS profiles at times 4/6 to 6/6). In contrast to the typical simplified geometry results, the higher WSS seen in the lateral sides of the TBJ during the expiration phase is notably absent in the scanned hummingbird simulation (see Figure 4.6: Top row: Mass flow rates over one breathing cycle. Velocity profiles (middle row) and corresponding WSS profiles (bottom row) of the hummingbird simulation at six selected times of the breathing profile, as illustrated in the graph in the top row. Inset figures A and B show a more detailed view of the WSS profiles near peak expiratory and inspiratory flow, respectively.)
Inset A in Fig. 4.6). This is caused by the constrictions corresponding to the vocal folds, which caused the flow to be directed medially, away from the lateral walls during inspiration (see velocity profiles 1/6 to 3/6 in Fig. 4.6). In addition, regions of alternating high and low WSS can be seen in the superior regions of the trachea, which correspond to tracheal and bronchial rings that slightly constrict the flow.

Secondary flow patterns during inspiration diverged from those of the simplified geometries (see Fig. 4.7). Similar to what was seen for the 15° and 90° simplified models (see Fig. 3.7), the

Figure 4.7: Secondary profiles during peak inspiratory flow for the 15° idealized hummingbird (top left), 90° idealized (top right), and scanned hummingbird (bottom).
A 15° simplified hummingbird model showed a lack of vortices inferior to the TBJ, while the 90° simplified hummingbird model did exhibit vortices. In contrast, although counter-clockwise velocity patterns were seen in the posterior and clockwise velocity patterns were seen in the anterior portions of the scanned model, it stops short of forming a full, 360° vortex, as flow near the posterior and anterior walls was flowing in the lateral-medial direction. Minimum WSS regions (see Fig. 4.8) were clustered in regions where the geometry bulged away from the central area of the bronchi or trachea, and shared some of the same areas of very low WSS in both expiration and inspiration. However, minimum WSS during both periods had a few unique regions that were on the downstream side of the vocal folds of whichever direction the flow was moving: more clusters of low WSS on the superior side of the vocal folds during expiration, and the inferior side during inspiration.

Visually, WSS was higher in the scanned geometries than the simplified models during peak inspiration and expiration, even discounting the regions of highest WSS near the vocal folds (see Fig. 4.9). A quantitative analysis of the average and maximum WSS over the entire cycle confirms this; average WSS was about three times as large in the scanned model, and maximum WSS was ten to twenty times as large (see Figure 4.10). While the domains over which average WSS was

![Figure 4.8: Regions of low WSS during peak expiratory flow (left) and inspiratory flow (right).](image)
evaluated were different in length (see Table 4.1), the visual depiction in Fig. 4.9 and quantitative representation in Fig. 4.10 combine to show that other factors were at play beyond a discrepancy in tracheal and bronchial lengths. Table 4.1 sheds light into the average WSS difference, with the discrepancy at the tracheal inlet being 20% larger in the simplified geometry, and at the bronchi being 90% and 87% larger in the simplified geometry. Speaking generally, larger diameters result in lower average WSS. It is worth noting that the maximum WSS, occurring in the scanned
geometry in the region of the vocal folds, would likely not be as high if vocal fold compliance
were taken into account; this prediction is based on a study of a human airway demonstrated that
accounting for compliance resulted in a significant WSS decrease [11].

4.9 Scanned Little Swift Comparative Results

A depiction of the little swift simulation at various states during one breathing cycle is shown
in Fig. 4.11. The little finch simulation had a region of WSS in the inferior region of the bronchi.
This contraction of the airway can be seen near the inferior part of both bronchi; however, the
bronchi on the right side had a more extreme, stenosis-like contraction. This location had the
highest WSS value at every moment over the entire cycle. A close-up of this region can be seen in
Fig. 4.11, Inset A. Like the hummingbird simulation (see Fig. 4.6), the little swift simulation
exhibited elevated WSS near the pessulus during inspiration; however, the values reached at the
pessulus were of a higher magnitude in the little swift simulation. The region of the pessulus was
more jagged in appearance in the little swift, which likely contributed to the higher values seen. It
is possible that the jagged values were due in part to the lower resolution of the little swift scan.
Similar to the hummingbird scanned geometry, the little swift scanned simulation did not exhibit elevated WSS near the lateral sides of the TBJ during expiration. This is likely due to irregularities in the little swift geometry which redirected the fastest airflow medially (see Fig. 4.11, velocity profiles 1/6 to 3/6).

Secondary flow profiles of the scanned simulation just downstream of the pessulus during inspiration showed a greater degree of similarity to the simplified little swift model (see Fig. 4.12) than did the scanned hummingbird profiles (see Fig. 4.7). A clear clockwise vortex was present in

Figure 4.11: Velocity profiles (middle row) and corresponding WSS profiles (bottom row) at the time instances marked in the breathing profile graph (top center). Inset A and B indicate regions of higher WSS.
the posterior region of the scanned geometry cross section, with a fainter counterclockwise vortex in the anterior region. Minimum WSS regions (see Fig. 4.13) followed similar trends as the hummingbird scanned simulation. Regions of low WSS occurred in locations where the airway bulged laterally, and tend to be on the downstream side of these bulges.

WSS during both inspiratory and expiratory flows appeared to be greater in magnitude throughout the scanned simulation than in the simplified simulation (see Fig. 4.14). Examining
quantitatively this observation for both average WSS and maximum WSS (see Fig. 4.15) confirms what is seen visually in Fig. 4.14. Maximum WSS during expiration reached approximately 25 times greater in the scanned simulation, while during inspiration maximum WSS was about eight times greater in the scanned simulation. Average WSS was about four times higher in the scanned simulation. Both maximum and average WSS differentials can be in part explained by differences in cross-sectional area, which was 84% greater in the trachea in the simplified model than in the scanned model (refer to Table 4.1). Part of the discrepancy in tracheal diameter seen both with the little swift and hummingbird geometries could stem from the allometric equations themselves, which had a lower bound of 27 g, which was just below and above the mass of the little swift and hummingbird, respectively. Equal flow rates in a smaller area will result in higher average WSS.
values. Contractions beyond the nominal diameter, such as near the inferior end of the bronchi or close to the vocal folds, contributed to higher maximum WSS values.

4.10 Conclusion

Scanned geometries of a hummingbird and a little swift were used in CFD simulations and compared to simulations that used simplified geometries. Secondary flow patterns in the scanned simulations just inferior to the pessulus showed some similarities to their simplified geometry counterparts, exhibiting clockwise swirls in the posterior region and counter-clockwise swirls in the anterior region of the cross section. Average WSS compared more closely between scanned and simplified simulations than did maximum WSS. Differences in both average and maximum WSS values were attributed in large part to discrepancies in observed diameters in the scanned simulations when compared with allometrically predicated values, as well as contractions in the scanned airways not present in the simplified geometries. A region of high WSS was noted near the pessulus during inspiration in both simplified and scanned geometries.
Future work in simplified geometries could model the contractions corresponding to the vocal folds, as well as match observed diameters rather than rely on the allometric equations, which should yield a more direct comparison of WSS and velocity patterns. Another study could better account for WSS in the regions corresponding to the vocal folds in the scanned geometries by replacing the stiff wall boundary condition with a compliant wall boundary condition.

The presence of relatively high WSS near the pessulus in both simplified and scanned simulations confirm that this region is an area of interest when looking at the possible role that WSS could have played in the development of the syrinx. However, the quantity and magnitude of the differences between the scanned and simplified geometries (as mentioned in the previous paragraph) suggest that simplified geometries are ultimately of limited value when examining WSS magnitudes and WSS profiles in present-day birds.
5 CONCLUSION

Computational fluid dynamic simulations were performed on simplified and scanned geometries of the avian respiratory system in order to examine WSS and velocity profiles near the TBJ. The simplified geometries featured high WSS near the pessulus during inspiration and high WSS just inferior to where the trachea first begins to expand into the bronchi during expiration, with the highest overall values being during inspiration. Maximum WSS values were found to generally be higher in larger bird size geometries, moderate angles, and breathing waveforms which had the highest inspiratory velocities. The scanned geometries also featured elevated WSS patterns near the pessulus during inspiration; however, the maximum WSS location for the scanned geometries was near where the airways were most narrow.

5.1 Wall Shear Stress and Velocity Patterns in Simplified Geometry Models of the Avian Airway

5.1.1 Conclusions and Contributions

Simplified geometries of the avian trachea and first bronchi which consisted of intersecting cylinders were created using a commercial CFD package. The size of the geometry, the branching angle between the bronchi, and the breathing waveform were individually varied to assess their impact on WSS and flow velocity patterns. Grid and time independence studies were performed for each geometric size using WSS at a point over one full breathing cycle. Particle image velocimetry was used to validate the results. Normalized velocity profiles taken across the right
bronchi for both the PIV and CFD results matched to less than 3% during inspiration on average, and within less than 2% on average during expiration when compared to the normalized maximum velocity.

The size of the geometries, as well as the breathing frequency and tidal volume, were varied according to allometric equations which related bird mass to these quantities. Geometries corresponding to bird masses of 3 g, 30 g, 300 g, 3,000 g, and 30,000 g were studied. Average WSS decreased with increasing mass. Maximum WSS during expiration generally followed this trend; however, maximum WSS during inspiration generally increased with increasing mass, which was caused by the jet of air entering the TBJ impinging more strongly on the pessulus for larger sizes. Secondary vortices perpendicular to the bronchi formed during inspiration for bird sizes 30 g and larger. The location of minimum WSS moved inferiorly with increasing mass, and for sizes larger than 300 g corresponded to locations at which reversed flow ended.

The branching angle for the branching angle simulations varied from 15° to 90°. Average WSS varied minimally across these simulations. Maximum WSS during inspiration was highest for the 40° angle, and lowest for the 15° angle. The size of the TBJ and the sharpness of the pessulus were two factors identified as contributing to the maximum WSS magnitude during inspiration for these simulations. Secondary vortices were not observed inferior to the pessulus for the 15° simulation, most likely due to the extended length of the TBJ, but were observed in the rest of the angle-varying simulations.

Three different breathing waveforms were tested. The biomimetic waveform was derived using the first six terms of a Fourier series taken of an ensemble average of a pressure signal acquired from a zebra finch at rest. The other two waveforms, the 3/1 ratio and the 1/1 ratio, featured constant velocity portions with expiration to inspiration time periods of 3 to 1 and 1 to 1,
respectively. WSS and velocity patterns were similar across these simulations. Maximum WSS was highest in the 3/1 ratio simulation due to it possessing the largest inspiratory velocity magnitude. Secondary vortices were not observed in the 1/1 ratio simulation due to lower inspiratory velocities, but were present in the 3/1 ratio and biomimetic simulations.

Unsteady effects were observed in the WSS magnitudes of the 3/1 and 1/1 ratio simulations and by comparing the average WSS values of the mass-varying simulations to the magnitude predicted by steady pipe flow with identical average velocity. The 3/1 and 1/1 ratio simulations exhibited peaks of average WSS magnitude when switching from transition portions to steady velocity portions, and in maximum WSS when switching from the inspiratory to expiratory transition period to steady expiratory flow. In contrast, the maximum WSS when switching to steady inspiratory breathing increased slowly to the steady-state value. This behavior was explained through the Womersley number, which predicts a more plug-like profile at higher Womersley numbers (higher Womersley numbers correspond to flows that oscillate more frequently or have larger characteristic lengths than lower Womersley numbers). In the mass-varying simulations, increasing Womersley numbers, caused by the increasing tracheal diameter, caused average WSS to depart further from the WSS magnitudes predicted by laminar pipe flow theory.

The results in this section were, to the author’s knowledge, the first time that WSS has been examined in the avian respiratory system over the range of parameters studied herein, as well as the first time that a breathing profile derived from pressure measurements in an avian lung had been used as boundary conditions for avian airway CFD simulations. High and low regions of WSS near the location of the syrinx and the conditions that maximize and minimize WSS near the syrinx quantified in this work will give direction to future studies involving the development and
advantages of the syringeal vocal folds, provided that WSS is determined to have played a role in said development and advantages.

5.1.2 Future Work

Suggested future work for the simplified geometries primarily consists of creating additional complexities that take into account constrictions corresponding to vocal fold-like structures, namely:

1. Adding a constriction to the airway (corresponding to sound-producing structures, such as vocal folds) to evaluate WSS at contracted conditions.
2. Assessing the impact of the 1/1 ratio of trachea to bronchial diameter assumption on WSS and flow patterns.
3. Adding compliance to the airway wall to examine effects of compliance on airway WSS in birds.
4. Simulating sound-producing structure motion to examine how the vibrating syrinx effects velocity and WSS distributions.
5. Adding more points to be verified for grid- and time- independence studies, or alternatively adding another metric, such as vorticity.
6. Validating the model using PIV at additional locations and under unsteady conditions.

5.2 Wall Shear Stress and Velocity Patterns in Scanned Models of the Avian Airway

5.2.1 Conclusions and Contributions

In order to assess how WSS and velocity patterns differed in real airway geometries as opposed to the simplified models, hummingbird and little swift scans were used in computational fluid dynamics simulations. The geometries were first acquired using CT scans, and then extracted
using computer software. Grid and time independence studies were performed on the new geometries. Simulations using simplified geometries were created with similar branching angles and identical allometrically-generated breathing patterns in order to most accurately assess how much the scanned geometries differed from simplified models that were as close as possible to the real bird airways, while still using the same assumptions with regards to geometry formation.

The scanned hummingbird geometry resulted in higher average and maximum WSS values than did the simplified hummingbird geometry. Highest WSS values throughout the breathing cycle occurred at the vocal folds, which constricted the flow. A region of relatively higher WSS was not present near the lateral section of the TBJ in the scanned geometry model during expiration. However, high WSS values were present near the pessulus during inspiration. While full vortices perpendicular to the bronchi were not present in the scanned hummingbird geometry, there was counter clockwise rotation present in the posterior region and clockwise rotation in the anterior region of the trachea just inferior to the pessulus during inspiration. Minimum WSS locations were clustered near the lateral downstream side of the vocal folds.

The little swift scanned geometry also had higher average and maximum WSS than the simplified geometry of the same size. Maximum WSS was located near a constriction in the airway in one of the trachea consistently over the entire breathing cycle. High WSS was noted near the pessulus during inspiration, but not near the lateral sides of the TBJ during expiration. Minimum WSS locations, like the scanned hummingbird geometry, tended to be congregated near the lateral downstream side of the vocal folds. Vortices perpendicular to the bronchi were present in both the scanned and simplified little swift geometries.

The results of this section were, to the best of the author’s knowledge, the first time that medical-grade imaging was used to define the geometry for avian airway CFD simulations. The
results demonstrated that while some similarities do exist between the scanned and simplified geometries, many differences in WSS magnitudes and flow features were also present. Additionally, differences in maximum and minimum WSS location between the hummingbird and little swift simulations indicated that the results for one bird species cannot necessarily be used to draw conclusions about another. Therefore, this work has demonstrated that species-specific geometry should be used to draw conclusions about WSS and flow features within that species’ respiratory system.

5.2.2 Future Work

Suggested future work related to the scanned geometries includes additional and larger bird simulations, as well as taking measures to reduce the effect of approximations used in the present work, namely:

1. Examining the respiratory system of a larger bird, such as a duck, to explore a wider range of mass variation of scanned geometries.

2. Adding compliance to the walls of the scanned geometries to better approximate WSS in real birds.

3. Modifying the simplified geometry to match the tracheal and bronchial diameters, as well as vocal fold contractions.

4. Obtaining a breathing pattern and geometry from a single bird to further reduce approximations taken with regard to breathing waveform shape, amplitude, and frequency.
REFERENCES


The velocity fields generated using particle image velocimetry as part of the effort to validate the computational model (see Sec. 2.7 and Figs. 2.12 and 2.14) required that a method be developed to extract a velocity profiles in order to quantitively compare the particle image velocimetry (PIV) and computational fluid dynamics (CFD) results. This appendix details the methodology used to extract velocity profiles from the PIV results, and approximate wall shear stress (WSS) values.

A.1 PIV Velocity Profile Extraction Methodology

In order to extract a velocity profile from the inspiratory PIV data, two images were first exported from the DaVis imaging software. The first image was a greyscale image of the average velocity field, with higher intensity values corresponding to higher velocity magnitudes. The second image featured the same content as the first, but additionally had one image of the particle field (e.g., a grayscale version of Fig. 2.11) overlaid in order to clearly discern the boundaries of the airway.

Both images were rotated clockwise 40° so that the right bronchus in each image ran vertically (see Fig. A.1, left image). The images were then imported into MATLAB. The intensity values between the red circles were first examined for the combination image in order to find the boundaries of the airway. The boundaries were taken to be the points where the intensity values
spiked, as shown in Fig. A.1 (right image). The results-only image (which did not include an overlaid image from one PIV image pair) was then used for the remainder of the analysis.

The image intensity values were taken for the results-only grayscale image and normalized with reference to the highest intensity value found along the profile. The pixel range was centered around the y-axis and normalized to fall between −0.5 and 0.5. The CFD data were taken using a line probe, and then normalized according to the highest velocity, and centered around the y-axis and normalized by the pixel range. Velocity values were normalized because matching Reynolds numbers should guarantee that the profiles have the same shape, but velocity would likely not be identical when differences exist in density, viscosity, and characteristic length. See Fig. 2.13 for the resulting velocity profile. The same process was repeated for the expiratory data, with Fig. 2.15 showing the result of this process.
A.2 Data Comparison Methods

Wall shear stress, as discussed in Sec. 1.3, is the force per unit area that a fluid moving relative to a surface imparts on the surface. While 3D velocity information would be required to fully assess WSS at a given location, because WSS at the points analyzed at the wall in Fig. A.1 reside on a symmetry plane upon which the velocity was taken, the assumption that the WSS magnitude was dominated by the in-plane velocity field should be true. In order to estimate the gradient of the velocity at the wall, a third-order curve was fit to one-fifth of the data points nearest to the wall (see Fig. A.2). As the PIV data predicted velocity even outside of where the walls were

![Figure A.2: CFD results (left column) and PIV results (right column) for the inspiratory (top row) and expiratory (bottom row) cases. Black solid lines depict the third-order fit along the closest 1/5 points next to the wall. The fit was then used to estimate WSS.](image-url)
(see Figs. 2.12 and 2.14), the curve fit was forced to have a zero velocity at the wall. In order to calculate average and maximum velocity values along the PIV versus CFD velocity profiles, the CFD profile, which had fewer points, had points added using a linear interpolation algorithm at the same radial locations as the PIV data.