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Quantification of Nasal Airflow Resistance In English Bulldogs Using Computed Tomography and Computational Fluid Dynamics

Eric T Hostnik, Brian A Scansen, Rachel Zielinski, and Samir N Ghadiali

Department of Veterinary Clinical Sciences (Hostnik, Scansen) and the Department of Biomedical Engineering (Zielinski, Ghadiali), The Ohio State University, Columbus, Ohio, USA.

Abstract

Stenotic nares, edematous intranasal turbinates, mucosal swelling, and an elongated, thickened soft palate are common sources of airflow resistance for dogs with brachycephalic airway syndrome. Surgery has focused on enlarging the nasal apertures and reducing tissue of the soft palate. However, objective measures to validate surgical efficacy are lacking. Twenty-one English bulldogs without previous surgery were recruited for this prospective, pilot study. Computed tomography was performed using conscious sedation and without endotracheal intubation using a 128 multi-detector computed tomography (MDCT) scanner. Raw MDCT data were rendered to create a three-dimensional surface mesh model by automatic segmentation of the air-filled nasal passage from the nares to the caudal soft palate. Three-dimensional surface models were used to construct computational fluid dynamic (CFD) models of nasal airflow resistance from the nares to the caudal aspect of the soft palate. The CFD models were used to simulate airflow in each dog and airway resistance varied widely with a median 36.46 (Pa/mm)/(L/s) and an interquartile range of 19.84 to 90.74 (Pa/mm)/(L/s). In 19/21 dogs, the rostral third of the nasal passage exhibited a larger airflow resistance than the caudal and middle regions of the nasal passage. In addition, CFD data indicated that overall measures of airflow resistance may significantly underestimate the maximum local resistance. We conclude that CFD models derived from nasal MDCT can quantify airway resistance in brachycephalic dogs. This methodology represents a novel approach to

CORRESPONDING AUTHOR: Eric T. Hostnik; erichostnik@gmail.com.

Dr. Scansen's current address is: Department of Clinical Sciences, Colorado State University, Fort Collins, Colorado, USA.

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LIST OF AUTHOR CONTRIBUTIONS

Category 1

- a. Conception and Design: Author name (s): Eric T Hostnik, Brian A Scansen, Samir N Ghadiali
- b. Acquisition of Data: Author name (s): Eric T Hostnik, Brian A Scansen, Rachel Zielinski, Samir N Ghadiali
- c. Analysis and Interpretation of Data: Eric T Hostnik, Brian A Scansen, Rachel Zielinski, Samir N Ghadiali

Category 2

- a. Drafting the Article: Author name (s): Eric T Hostnik, Brian A Scansen, Samir N Ghadiali
- b. Revising Article for Intellectual Content: Eric T Hostnik, Brian A Scansen, Rachel Zielinski, Samir N Ghadiali

Category 3

- a. Final Approval of the Completed Article: Eric T Hostnik, Brian A Scansen, Rachel Zielinski, Samir N Ghadiali

noninvasively quantify airflow resistance and may have utility for objectively studying surgical interventions in canine brachycephalic airway syndrome.

Keywords

Brachycephalic airway syndrome; Airway resistance; Canine

INTRODUCTION

The English bulldog breed is growing in popularity.¹ However, the breed is predisposed to multiple congenital disorders that may have a negative impact to the health of the dog. Brachycephalic airway syndrome (BAS) is a collection of respiratory structural anomalies of the respiratory tract. Each anomaly contributes to an overall compromise of airway patency. The abnormalities span from the nares to the distal trachea.² The severity of abnormalities for an individual dog is difficult to objectively assess using only physical examination and laryngeal visualization. Brachycephalic breeds like English bulldogs may have few clinical signs of airway disease; however, the spectrum of affliction is wide and more severely brachycephalic dogs may rapidly develop respiratory distress with the potential of death.³ Surgery is performed to address conformational obstruction with the goal to decrease the airflow impedance and reduce the upper airway resistance.⁴ There is no quantitative tool to objectively assess if a reduction in airflow resistance occurs post-surgery.

Shortened craniofacial skull conformation provides insufficient space necessary for appropriate intranasal development.⁵ The abnormal shape of the nasal cavity is thought to develop in an incongruent (heterochronic) growth pattern with truncation of the bones at the base of the skull, as well as the exterior facial bones, while the nasal/ethmoid turbinates continue to develop normally.⁶ The consequence is a nasal cavity filled with many convoluted, intricate scrolls of bone and overlying congested mucosal tissue.⁷ The intertwining of structures causes increased mucosal contact points that decrease intranasal passageways with a relative excess of nasal turbinates.⁸ The lack of space for growth of these nasal bones culminates in aberrant rostral and caudal turbinates that occupy the airway passage, which may further potentiate airflow impedance.^{3, 9} Brachycephalic airway syndrome is used to summarize the effects of the anatomic anomalies that negatively impact the airway.¹⁰ Despite a high prevalence and wide spectrum of affliction within brachycephalic breeds, there is a lack of quantitative measures to help characterize how severely individual dogs may be affected.^{11, 12} Previous work has used a combination of video-endoscopy and barometric whole-body plethysmography to evaluate brachycephalic airway disease with no significant correlation to severity of clinical signs.^{13, 14}

Brachycephalic airway syndrome has classically included stenotic nares (Figure 1A), elongated soft palate with excessive soft palate tissue (Figure 1B), laryngeal collapse, everted laryngeal sacculles, and a hypoplastic trachea.^{15, 16, 17, 18} More recent literature has investigated additional factors that may contribute to the impedance of airflow; these include increased mucosal contact points of the nasal turbinates, nasopharyngeal turbinates (caudal aberrant turbinates), rostral aberrant turbinates, macroglossia, and amygdalitis (Figure 1C,

Figure 1D).^{6, 9, 19} Excess tissue obscures an already highly restrictive airway leading to a further increase of resistance. The nasal cavity contributes 76.5% of the total airflow resistance in non-brachycephalic dogs with the larynx and bronchi contributing 4.5% and 19%, respectively.²⁰ The nasal cavity resistance increases to 80% in brachycephalic dogs.²⁰ This rostral increase of resistant airway dynamics is similar to humans, especially people afflicted with nasal airway obstruction (NAO) secondary to complex nasal defects.²¹ The increased upper airway resistance of brachycephalic dogs has also been correlated to an increased rate of gastrointestinal abnormalities like gastroesophageal reflux disease.⁴

When activity restriction is not sufficient, surgical intervention is a keystone treatment for brachycephalic airway syndrome dogs.¹⁵ Surgical techniques may include alar wedge resection or alarplasty, partial staphylectomy, and laryngeal ventriculectomy.^{22, 23, 24} Overall, the prognosis for surgery is good with low perioperative mortality; however, there is little explanation for why some dogs do not respond well to surgery.^{4, 25, 26}

Computational fluid dynamics (CFD) has recently become a tool to aid surgical planning in humans with nasal airway obstruction.^{27, 28} Personalized models of individuals have been created from CT images to test mechanics of flow through an *ex vivo* model to aid surgical planning. Studies have shown that this tactic may be used before surgery and then following surgery to quantitatively assess functional outcomes of surgical intervention.^{29, 30}

Computational fluid dynamics has been used to evaluate air flow through the glottis in cadaveric models of Thoroughbred horses, as well as assessing the effect of airflow for olfaction in dogs.^{31, 32, 33, 34, 35} Although similar computational techniques have been used to evaluate otolaryngological disorders in humans, to the authors' knowledge, application of this technology to brachycephalic airway syndrome is a novel approach to evaluating upper airway resistance in veterinary medicine.³⁶

Computed tomography allows for screening of subclinical lower airway pathology while simultaneously evaluating upper airway structures. A computed tomographic scanner with a high number of detectors minimizes the time required for imaging. Longer acquisition times necessitate general anesthesia to facilitate that a patient is compliant enough for a diagnostic quality study. The combination of gastroesophageal reflux disease and prolonged anesthesia increases the risk to develop aspiration pneumonia, especially in brachycephalic breeds.^{4, 37, 38} Therefore, one goal of this study is to demonstrate that the use of a 128-detector multi-detector computed tomography (MDCT) scanner can rapidly acquire diagnostic quality images without the need of general anesthesia to assess brachycephalic airway syndrome in English bulldogs. We also seek to demonstrate that these high quality MDCT images can be used to generate patient-specific computational fluid dynamic models of the nasal passages in English bulldogs and demonstrate that this computational technique represents a novel way to quantify and characterize airway resistance through the nasal cavity.

METHODS

The study was a prospective, pilot design. Twenty-one privately owned English bulldogs were recruited for this study from clients of The Ohio State University Veterinary Medical

Center. English bulldogs were recruited due to the prevalence of brachycephalic airway syndrome within the breed and an attempt to standardize the computational fluid dynamic technique. The sample size was maximized to the available funds in order to improve the power of statistical analysis. Medical history for each patient was obtained. The dogs underwent routine physical examination, cardiovascular examination that included auscultation of the heart with palpation of peripheral pulses by a board-certified veterinary cardiologist, and a screening echocardiogram. Dogs were excluded from the study if they were determined to be a risk for a sedated CT study. Dogs were also excluded if there was previous history of airway surgery or facial trauma. A board-certified veterinary cardiologist with an interest in airway disease of brachycephalic breeds (B.A.S.) made decisions for subject inclusion or exclusion. Informed client consent was obtained for all dogs and the study was approved by the Institutional Animal Care and Use Committee of The Ohio State University as well as the Clinical Research and Teaching Advising Committee of The Ohio State University Veterinary Medical Center.

All computed tomography (CT) examinations were performed using conscious sedation and dogs were not intubated. Dogs received an intramuscular injection of butorphanol (0.20 mg/kg) and dexmedetomidine (5 to 10 mcg/kg) 30 to 45 minutes prior to the scheduled CT scan. A cephalic venous catheter was placed to maintain vascular access and standard anesthetic rate of intravenous fluids were provided during the imaging. Atropine (0.04 mg/kg) or glycopyrrolate (0.01 mg/kg) were given on an individual case by case basis if clinically-significant bradycardia developed, as determined by a board-certified veterinary cardiologist. Anesthetic monitoring (electrocardiography, pulse oximetry) was performed throughout the imaging. Dogs were under direct supervision until deemed adequately recovered from sedation.

The dogs were positioned in sternal recumbency with the neck extended, the forelimbs slightly abducted, and the hard palate parallel to the CT table with table straps to help secure the dog in place. A dual source, 128-slice, multi-detector CT scanner (Somatom Definition Flash, Siemens Healthcare, Malvern, PA), with temporal resolution of 75 ms and maximum scanning speed of 458 mm/s, was used to perform the scans. The gantry angle was set to 0 degrees. Topograms were acquired in laterolateral and dorsoventral views to plan the slice series. Scan parameters were carried out with a fluctuating filament current that had a maximum milliamperage of 495 mA and fluctuating tube current of 100 – 120 kVp. The region of interest was scouted to include rostral to the nasal planum through the caudal aspect of the diaphragm. The CT dataset was reformatted with soft tissue and bone algorithms with appropriate window level and window width Hounsfield units to optimize conspicuity of upper airway structures. The slice thickness was acquired at a maximum of 1 mm. The spiral pitch varied from 0.17 to 0.65 based on the electrocardiogram as the dogs were also imaged using cardiac-gating for a second study focused on cardiac angiography. Intravenous contrast was not administered for the airway portion of the study. The CT dataset was reformatted into dorsal, transverse, and sagittal planes with the ability to manipulate multiplanar reformats. The datasets were saved as Digital Imaging and Communications in Medicine (DICOM) files and studies were stored on the local picture archiving and communication system (PACS).

The processes used for model construction and meshing for computational fluid dynamic analysis are shown in Figure 2. All procedures were performed by a third year radiology resident (E.T.H.) under the guidance of laboratory personnel with a focus on airway based and Eustachian tube based computational fluid dynamics (R.Z and S.N.G.). First, the raw MDCT bone algorithm transverse dataset was imported into a three-dimensional (3D) image processing and model generation software program (ScanIP, Synopsys, Inc. Simpleware, Version 7.0, Chantilly, VA). The anatomy imported into the 3D image processing software included the entire nasal passage including frontal sinuses from nasal planum to the laryngopharynx. The rostral border was cropped at the first image in which the nasal apertures, at the confluence of the wing of the nostril, were closed off to the external surroundings. A threshold was applied to the CT layers by automatic segmentation using -1024 to -450 Hounsfield units (HU) to highlight the airways (blue in Figure 2B). A fill threshold was then applied to isolate the air-filled nasal passage throughout contiguous images which generated a 3D solid model (Figure 2C). Still within the 3D image processing program, the solid model was sub-divided into a computational fluid dynamic mesh consisting of 6-sided tetrahedral elements which was then exported as a matrix laboratory (MATLAB) file into a finite element analysis, solver and simulation software package (COMSOL, Multiphysics 5.0, COMSOL, Inc., Version 5.0.1.276, Burlington, MA). During this meshing procedure, we specified the minimum edge length which is defined as the minimum length of any side of all tetrahedral elements in the model. As a result, the density of the mesh (and hence the number of elements) could be controlled by specifying the minimum edge length where smaller minimum edge lengths lead to a more dense mesh with larger number of elements. For this study the minimum edge length was set to 0.3 mm and if a three-dimensional (3D) mesh was not able to be constructed using 0.3 mm edge lengths, mesh construction was reattempted by increasing the minimum edge length at 0.05 mm increments.

A finite element algorithm was then used to simulate airflow in each 3D model of the nasal passage and to calculate airway resistance. For this analysis, the airways were filled with an incompressible, Newtonian fluid with constant density and viscosity that mimicked the properties of air at environmental conditions. Iterations of the Navier-Stokes equation for incompressible flow were used to solve the momentum balance and continuity equations governing airflow³⁹

$$\rho \left(\frac{\partial \vec{u}}{\partial t} + (\vec{u} * \nabla) \vec{u} \right) = -\nabla p' + \mu \nabla * (\nabla \vec{u} + (\nabla \vec{u})^T) + \rho \vec{g} \quad (1)$$

where ρ is the density, $\frac{\partial}{\partial t}$ is a partial derivative, $\vec{u}$ is the fluid velocity vector, p' is the pressure, μ is the dynamic viscosity, ∇ is the spatial gradient operator, and $\vec{g}$ is the acceleration due to gravity. Boundary conditions of no flow on airway walls, zero reference pressure at the nares and an airflow rate of 0.5 L/s at the most distal airway section near the nasopharynx were used to solve Equation 1 in dimensional formats. The primary output was the airflow velocity field and the distribution of pressure within the nasal cavity. Airflow had a laminar velocity at the inlet with an atmospheric pressure of zero assigned to the nares.

The properties of the airflow were maintained between all of the dogs. Airway pressure maps were generated as color-coded overlays representing the shape of the upper airway anatomy (Figure 3). Pressures values were determined as a static value from the surface average at 5% equidistant intervals throughout the airway (Figure 4). Flow using the Navier-Stokes equation is driven by high pressure at the inlet to a lower reference pressure (zero) at a determined outlet. The airway model was divided into percentages of 5% increments from rostral to caudal for purposes of measuring airway pressure. The zero pressure was set at the 0% slice located at the nares. The 0% slice was determined to be the point that the nasal passage formed a complete border. The 100% slice was set at the most caudal aspect of the soft palate just rostral to the confluence of the nasopharynx and oropharynx into the laryngopharynx. The most caudal slice therefore has the highest pressure due to the rise and build up of pressure through the nasal passage. The pressure changes along the airway lead to airflow within the airway which is demonstrated by the black arrows in Figure 3.

$$Rv = \frac{\Delta P / L}{Q} \quad (2)$$

$$Rv = \frac{dP}{dx} / Q \quad (3)$$

The overall resistance in the nasal passage was calculated using Equation 2 with P set to the overall pressure change from nares to the most caudal slice measured in (Pa/mm)/(L/s), L is the total length of the nasal passage measured in mm and Q is the airflow rate set to 0.5 L/s. In addition, the local resistance was calculated using Equation 3 where dP/dx represents the spatial derivative of pressure between the 5% increments (Equation 3). A greater degree of pressure change is demonstrated by a change in color for the color flow maps, as well as a greater slope on the line graph (Figure 5). The greater change in pressure and resistance had a higher slope as quantified by Equation 3. We chose to divide the airway into thirds when assessing the sites of greatest airway resistance. The rostral one-third includes the nasal planum to the rostral aspect of the frontal sinus. The middle one-third includes the frontal sinus into the choanae ending at the pterygoid bone. The caudal one-third extends from the caudal pterygoid bone/hard palate to the confluence of the nasopharyngeal meatus/soft palate to the laryngopharynx.

The repeatability of calculating airway resistance was evaluated in three English bulldogs. The repeatability was assessed by regenerating the mesh structures from the original DICOM files, formulating the model, and then solving the computational fluid dynamics equations. The generation of a second 3D mesh was repeated with the same procedure using identical parameters within the 3D image analysis software program starting from contiguous transverse CT DICOM files. The mesh was then exported as a MATLAB file, imported into the finite element analysis software package, and computational fluid dynamic modeling was performed a second time. The repeatability of airway resistance calculation was calculated by determining the percent difference between the two results yielded.

Statistical analysis was performed by the authors (E.T.H., B.A.S., S.N.G.) using commercial software (SPSS Software, Version 22.0.0, IBM Corp, Armonk, NY). Descriptive statistics including the median and standard deviation were calculated for the airway resistance by operator and examined for normality using inspection of scatterplots and the Kolmogorov-Smirnov test. Means of the age and weight were calculated. Effects of age and weight on airway resistance were assessed by linear regression. Friedman repeated measures analysis of variance (ANOVA) was used to compare the maximum airway resistance to the overall resistance, as well as comparing between the three regions for maximum resistance.

RESULTS

Twenty-one English bulldogs without previous history of airway surgery or cranial trauma were enrolled in the study. None of the dogs were excluded from further analyses. The average age was 28.5 months (range of 2.9 to 108.6 months). Four castrated males, eight intact males, four spayed females, and five intact females were enrolled. Computational fluid dynamic models were successfully constructed for all animals and as shown in Figures 3 and 4, simulations resulted in airflow and pressure distributions within the nasal passage. These pressure distributions were used to evaluate both the overall and local resistance to airflow in each dog using equations (2) and (3).

Prior to data analysis, a mesh refinement analysis was conducted to evaluate the accuracy of the numerical solution. Specifically, the solid geometry for one dog was meshed using different minimum edge length values and this resulted in models with a different number of tetrahedral mesh elements. Numerical simulations were then performed in each model to calculate the overall and maximum resistance. Note that in this analysis, the mesh density was inversely proportional to the minimum edge length. As shown in Figure 3B, increasing the mesh density resulted in a decrease in the overall and maximum resistance. However, for minimum edge lengths less than 0.5 mm, or inverse minimum edge length greater than 2 mm^{-1} , the overall and maximum resistance plateaued to values that were within 10%. Therefore, mesh densities with minimum edge length greater than 0.5 mm results in mesh-independent solutions. Note that in this study we utilized minimum edge lengths of 0.3 mm since computational resources allowed for efficient calculation at this mesh density.

The distribution of the pressure and resistance data set was non-normal and airway resistance varied widely with a median overall resistance of 36.46 (Pa/mm)/(L/s) and an interquartile range of 19.84 to 90.74 (Pa/mm)/(L/s). Airway resistance did not correlate with age ($r = 0.344$, $P = 0.126$) or weight ($r = -0.058$, $P = 0.803$). The repeatability data were variable with the second calculation of airway pressure determined to be 79.0%, 95.3%, and 93.2% similar to the initial calculation for the three dogs in which repeatability was tested.

The increase in pressure was not uniform through the airway (Figure 5). Local resistance (which is approximately the slope shown in Figure 5) was not homogeneous throughout the airway as pressure changes varied with respect to distance along the airway. As a result, the maximum local resistance was statistically larger than the overall resistance (Figure 6A). In addition, the greatest step up of pressure occurred within the rostral one-third of the airway in 19 of 21 dogs (Figure 5) and as a result the maximum local resistance within the rostral

third of the nasal passage was significantly higher than the maximum flow resistance in the middle third and caudal third segments (Figures 6B). A representative model of pressure change within the airway anatomy correlated the largest impedance of flow to the anatomy from the nasal planum to the caudal nasopharynx, which consists primarily of the nasal turbinates (Figure 5). The remaining two dogs had the largest local increase of airway resistance occur in the caudal one-third of the airway, at the confluence of the nasopharynx into the laryngopharynx (Figure 5). Asymmetry of airway pressure between left and right nasal cavities was also identified in two dogs (Figure 7).

DISCUSSION

Findings from our study indicated that numeric measures of airway resistance are achievable to quantify upper airway obstruction in brachycephalic dogs. Twenty-one English bulldogs successfully had computational fluid dynamics applied to a 3D finite element model construction derived from MDCT studies. To the author's knowledge, computational fluid dynamics have not been previously used in the brachycephalic dog to map, model, and measure pressure changes within the nasal passage. The color map and characteristic curve demonstrates the change of pressure through the nasal passage (Figure 5). The anatomy with color map correlates the impedance of flow with structures throughout the upper airway. A wide degree of variation for the quantified resistance was observed between the 21 English bulldogs within this study. Similarly, clients reported differing severity of clinical signs ranging from mild to moderate and it is likely that this population of English bulldogs were not clinically affected by to the same degree, supporting the notion that brachycephalic airway syndrome is a spectrum.

Mapping changes of airway pressure through the model is a promising tool for addressing management of disease. Current surgical intervention to address airway obstructive disease includes widening the nasal apertures and removal of redundant tissue within the caudal pharynx caused by the soft palate. Our data showed that the rate of change in local pressures throughout the nasal passage is not uniform and anatomy is likely one of the major contributions to the inhomogeneity. If the rise in pressure in the airway was uniform, then the slope of the change in resistance from the nares to the caudal nasopharynx would be constant and represent a linear function. However, maximum local resistance value is significantly greater than the overall resistance (Figure 6). Our data also localizes the greatest step-up of airway resistance to occur within the rostral third of the upper airway in 19/21 English bulldogs, which includes not only the nasal apertures but also the nasal turbinates with their edematous mucosa. This has important clinical implications as it suggests the most important area to target for surgical widening of the nasal passages. Focusing the surgical effort to only the nares and nasal cavity without soft palate intervention would be desirable as it would reduce anesthesia time and surgical duration without having to manage post-operative inflammation in an already stenotic nasopharyngeal/laryngopharyngeal area.

Edematous mucosa has been proposed as a complicating factor to airway disease, but surgical correction in this area has not become routine. Nasal turbinectomy using a diode laser has been proposed and performed, but this intervention has yet to become common

practice.^{40, 41} Based on our modeling results, nasal turbinectomy could theoretically alter the anatomy found to contribute to the highest region of airway resistance in the 19 of 21 English bulldogs. A turbinectomy would initially increase the diameter of the nasal passages resulting in a decreased pressure according to Poiseuille's law. Reported complications observed after nasal turbinectomy have included regrowth and hyperplasia of the turbinates in between 65%–98% of brachycephalic breeds, which negated the effectiveness of the procedure.⁴² Regrowth of nasal turbinates with recurrent increased airway resistance is also seen in people after turbinectomy.⁴³

Airway pressures between the left and right nasal passages differed in two dogs of the current study (Figure 7). Asymmetry in airway pressures between nasal cavities of an individual dog may or may not be a consequence of deviation of the nasal septum; septal deviation has been identified in 21% of brachycephalic dogs in a post-mortem study.¹⁹ With these data, surgical intervention may be directed to alleviate pressure within the more affected side to minimize trauma caused by surgery. Directing efforts to the region of the anatomy contributing to the greatest change in pressure would maximize the impact of surgery. However, a physiologic phenomenon consisting of cyclical congestion and decongestion of the venous sinusoids of the nasal mucosa referred to as nasal cycling may also contribute to airway asymmetry in CT studies.⁴⁴ Nasal cycling was not screened for or controlled in the current study.

This study developed a model that included the nasal airway from the nasal planum to the caudal soft palate. The model was directed to encompass anatomy addressed by surgery and was limited to the portion of the airway that had defined circumferential borders. As such, the model excluded other components of brachycephalic airway syndrome such as hypoplastic trachea, everted laryngeal sacculles, or laryngeal collapse.^{45, 46} The model also excluded the dynamic component of brachycephalic airway syndrome, which may include laryngeal dysfunction and intermittent obstruction of flow from the soft palate. Decreasing the overall radius of a tubular system such as the nasopharynx will increase the overall resistance to the fourth degree, according to Poiseuille's Law.⁴⁷ The chronic increased resistance has been theorized to culminate in terminal stage airway disease of laryngeal collapse.¹² No dogs investigated in this study had everted laryngeal sacculles or laryngeal collapse.

The repeatability was more variable than desired; however, the dog with a repeatability value of 79.0% was generated from a dog imaged early in the study, whereas the 95.3% and 93.2% were dogs from the latter part of the study. This may indicate a learning curve, as well as recognizing the value of consistently imaging at a smaller spatial resolution.

Limitations of this study include lack of a gold standard for airway resistance and the variable clinical severity of the dogs enrolled. While values for airway resistance were obtainable, repeatable, and were logically correlated to anatomic sites of airway narrowing in dogs with brachycephalic airway syndrome, the accuracy of the model and computational fluid dynamic data cannot be proven.

In conclusion, findings from this pilot study indicated that computational fluid dynamics derived from nasal MDCT appear able to quantify airway resistance in English bulldogs. This methodology is a new approach to measure airway resistance and may have utility for objectively studying surgical interventions in canine brachycephalic airway syndrome. A study measuring the airway resistance and clinical signs following nasal surgery is needed to evaluate the effectiveness of surgical intervention on the nasal passage alone. Future studies will include the use of this methodology to quantitatively assess dogs with severe brachycephalic airway syndrome pre- and post-surgery.

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Figure 1.

Computed tomography of brachycephalic airway syndrome: stenotic nares (A), elongated soft palate (B), and caudal aberrant nasopharyngeal turbinates (C & D). A. Transverse reformat at the rostral nasal planum. White arrows show stenotic nares. B. Sagittal reformat on midline. White arrow shows thickened/elongated soft palate. C & D. Transverse reformat at the choanae and sagittal reformat on midline. White arrows show caudal aberrant turbinates. Dog is in ventral recumbency with the right side of the dog in the left aspect of the image or rostral to the left. Bone algorithm. Window width: 2500 Window level: 250.

Figure 2.

The process for generation of a 3D model. A: Computed tomographic images of the nasal passages were acquired in a transverse image plane using multi-detector computed tomography. A representative image of the caudal nasal passage at the level of the choanae is shown. B: Threshold segmentation in the ScanIP program isolated the airways (blue) using a threshold of -1024 to -450 Hounsfield units (HU). C: A fill threshold algorithm in ScanIP was used to generate a solid model of the nasal passages which was then sub-divided into 6-sided tetrahedral elements and exported to the COMSOL Multiphysics software package. D: 3D model of the nasal passages within the COMSOL Multiphysics package with E: Zoomed in image of the 3D model and tetrahedral mesh elements used in the computational fluid dynamic analysis. Note that each tetrahedral element has 6 sides or edges and the minimum edge length controls the mesh density where a smaller minimum edge length leads to a more dense mesh and more elements.

Figure 3.

A: COMSOL Multiphysics 5.0 with MATLAB® model of pressure generated using computational fluid dynamics. The scale on the right is a relative scale displaying pressure with measurements in Pascals. Low-pressure areas are blue and high-pressure areas are dark red. The areas with a transition in color represents an increase in pressure within the model. The black arrows within the model are a logarithmic representation of velocity of air flow in m/s. A larger arrow represents a higher velocity. The arrows are directed to the nares reflecting the pressure drop that correlates to the zero reference for the Navier-Stokes equation. The orientation of anatomy is the same as the positioning for Figures 1B and 1D with the rostrum to the left of the image. The z-, y-, and x-axes refer to the imaging planes of computed tomography. B: Pressure distributions were used to calculate overall and maximum airflow resistance at different mesh densities as characterized by the inverse minimum element edge length. Data indicate that mesh densities with minimum edge length less than 0.5 mm (or inverse minimum edge length greater than 2 mm^{-1}) resulted in solution values that were within 10% and are thus mesh-independent solutions.

Figure 4.

Computational fluid dynamics model in 20 equally distributed slices of the same dog as Figure 3 (A: lateral, B: rostrocaudal). The rostral airway is to the left. The scales display pressure measurements in Pascals; low pressure is blue and high pressure is dark red. A mean of the surface pressure for each slice was measured at 5% increments from the nares to the caudal nasopharynx.

Figure 5.

Line graph representing the change of pressure relative to the nasal airway anatomy. The scale along the right side represents a percentage of the total airway resistance. The dashed black line is the overall average change in pressure for all 21 English bulldogs. Each solid line represents one dog. In 19/21 dogs, the greatest step up of pressure was within the rostral one-third of the airway. The black arrows within the airway are a logarithmic representation of velocity of air flow in m/s. A larger arrow represents a higher velocity.

Figure 6.

A: Paired measurements of overall and maximum flow resistance in 21 dogs. Measurements within each subject is shown via common symbol and solid connecting line. Dashed line represents median of 21 dogs. B: Measurements of maximum flow resistance in the Rostral 3rd, Middle 3rd, and Caudal 3rd of the nasal passages. Data is shown with standard box-whisker plots where the median is the horizontal line. * indicates significant differences via Friedman Repeated Measures ANOVA.

Figure 7.

Asymmetry of pressure between the right and left nasal passages was identified in two dogs.

A: Rostrocaudal orientation; smoothed 3D model of the airway showing the right nasal cavity having a higher pressure compared to the left nasal passage. The yellow color depicts a higher pressure relative to the blue color-coded left nasal cavity. B: Ventrodorsal orientation; smoothed 3D model airway shows the differing pressures within the rostral nasal passages. The rostral nares are at the top of the image. C: Transverse CT image of a nasal cavity with asymmetry between the nasal cavities. Dog is in ventral (V) recumbency, the right side (R) of the dog is to the left of the image. Bone algorithm. Window width: 2500 Window level: 250.