Pulsatile flow in the nasal cavity with high flow therapy measured by stereoscopic PIV

Callum J. T. Spence¹, Nicolas A. Buchmann¹, Mark C. Jermy¹

1: Dept. of Mechanical Engineering, University of Canterbury, Christchurch, New Zealand, callum.spence@pg.canterbury.ac.nz

Abstract Nasal high flow (NHF) cannulae are used to deliver heated and humidified air to patients at steady flows ranging from 5-50 l/min. In this study, the flow velocities in the nasal cavity across the complete respiratory cycle during natural breathing and with NHF flow have been mapped in-vitro using time resolved stereoscopic particle image velocimetry (SPIV). An anatomically accurate silicone resin model of a human nasal cavity was constructed using CT scan data and rapid prototyping. Physiological breathing waveforms were reproduced in-vitro using Reynolds and Womersley number matching. As well as contributing to the understanding of the airflow through the nasal cavity during natural breathing, these measurements also help to elucidate the effect of assisted breathing on respiratory mechanics and patient comfort. The flow pattern in the nasal cavity with NHF was found to differ significantly from natural breathing. Preliminary results have been shown that suggest a quasi-steady flow assumption for natural breathing is inaccurate, however, time dependency was not found with NHF.

1. Introduction

The primary function of the nasal cavity is to filter, warm and humidify inspired air before it reaches the lungs, while providing olfactory function. All these functions are dependant on the flow field in the nasal cavity. Nasal cannulae are used to administer a breathing ventilation therapy known as nasal high flow (NHF). Cannulae therapy until recently has been limited to low flow rates due to the discomfort and irritation caused by delivering dry, cold gas to the nasal passages [1]. NHF, however, delivers heated and humidified air at body temperature pressure saturated (BTPS) to patients at steady flows ranging from 5–50 l/min via a nasal cannula.

In this study, the flow velocities in the nasal cavity with and without NHF flow have been mapped in-vitro using stereoscopic particle image velocimetry (SPIV). As well as contributing to the understanding of the airflow through the nasal cavity during natural breathing, these measurements also help to elucidate the effect of assisted breathing on respiratory mechanics and patient comfort. In order for clinicians to make informed decisions on NHF flow rates, it is necessary to understand the effect of nasal cannula airflow on the flow velocity distribution and pressures in the nasal cavity.

Numerous experimental and computational flow studies have been undertaken in the nasal cavity. Doorly et al [2] and Chung et al. [3] have provided recent reviews on the mechanics of airflow and the use of PIV in the nasal airways. Most measurements conducted to date have assumed quasi-steady conditions in the nasal cavity based on a small Womersley number for quiet breathing of around 3; however, uncertainty still remains whether the flow can be modeled as steady. Horschler et al [4] and Lee et al [5] both describe major differences between steady and unsteady flow simulations, particularly during expiration and near the transition between inspiration and expiration. This study, to the authors' knowledge presents the first SPIV measurements conducted in a complete nasal cavity model with pulsatile flow, both with and without NHF.
2. Experimental Method

2.1 Nasal cavity flow phantom

A 1.55 times scale flow phantom of a complete human nasal cavity was constructed from computed tomography (CT) data. The CT data set of a 44-year-old male comprised 452 axially acquired 512x512 pixel resolution images with a 0.6 mm slice spacing and thickness. A radiologist declared the nasal cavity free of any visible abnormalities. A circular cross-section was lofted to the termination of the larynx to facilitate connection of the model to flow conduits. The geometry was rapid prototyped (Figure 1.a) in a water dissolvable material and embedded in a clear silicone resin. After the silicone resin had cured the model was removed, leaving a transparent flow phantom of the nasal cavity (Figure 1.b). Both sides of the nasal cavity and the lofted trachea termination are visible in Figure 1.c.

![Figure 1. (a) Nasal cavity rapid prototyped geometry and (b) silicone nasal cavity flow phantom viewed sagittally from the left and (c) axially from the bottom.](image)

The nasal valve occurs just posterior of the nasal vestibule and is the region of smallest cross-sectional area. The nasal valves for the in-vitro model used in this study can be seen as the curve minima labeled 1L and 1R in Figure 2 and have cross-sectional areas of 1.11 and 1.15 cm$^2$ respectively. These cross-sectional areas lie within the 0.54–1.21 cm$^2$ range measured by Çakmak et al [6] using CT data from 25 healthy adults. Following the nasal valve is an abrupt expansion into the main cavity both in height and cross-sectional area. Although each side of the nasal cavity share common features they are asymmetric. The cross-sectional area of the right nasal cavity 7.2 cm posterior of the nostrils was 30% larger than on the left. Beyond the nasal cavity the nasopharynx narrows to a minimum cross-sectional area of 1.1 cm$^2$; notably smaller than the combined cross-sections of the two nasal valves.

The shape of the external nose influences the flow entering the nostrils and was therefore included in the phantom. Physiological features such as nostril hairs and mucous membranes inside the nasal cavity were not resolved because the main interest of this paper was the large and medium scale flow features, which were assumed independent of these details. Similarly, and to reduce optical noise, the sinuses were not resolved. The cannula was rapid prototyped in clear stereo-lithography resin with a refractive index of 1.51. Reflections from the cannula were, however, minimised by painting the cannula models matt black.
2.2 Flow System

The nasal cavity model was installed in a recirculating flow system as shown in Figure 3. A mixture of water and glycerine was used as a working fluid to match the refractive index of the flow phantom. The refractive index of the silicone rubber used for the phantom construction is specified as 1.43 by the manufacturer, however, varies due to differences in mixing and curing from model to model. The optimum mixture for this study's model was found to be 39% water and 61% glycerine by volume. The liquid temperature was maintained constant at 32°C with a cooling and heating system. The viscosity was measured using a Haake RV20/RC20 viscometer to be $7.3 \times 10^{-3}$ Pas and the density determined to be 1140 kg/m$^3$. The viscosity and density of in-vivo BTPS air are $18.8 \times 10^{-6}$ Pas and 1.068 kg/m$^3$ respectively.

A reciprocating piston pump driven by a ball screw and stepper motor was constructed to provide a physiological breath waveform in-vitro. Breathing waveforms were measured in-vivo during natural breathing and with NHF on a healthy 23 year-old male with height 184 cm and weight 85 kg. Average waveforms (Figure 4) were created from 20 breaths and were fitted to a Fourier series. The amplitude of the waveform was scaled using Equation 1 to match the in-vivo and in-vitro Reynolds
numbers. A flowrate of 30 l/min in-vivo corresponded to an in-vitro flow rate of 16.9 l/min. Similarly, the period, T, was scaled to maintain a constant Womersley number by Equation 2. A breath period of 5.4 s in-vivo corresponded to an in-vitro period of 35.7 s. A constant pressure header tank supplied steady flow to the cannula, which was measured with an electromagnetic flow meter (Tigermag FM626). Return and overflow lines connected the reservoir and a header tank to ensure flow over the weir and constant head. The phantom was immersed in the reservoir to a depth such that the free surface remained flat at the highest flow rate, effectively providing a constant pressure at the nostrils.

\[
Q_{\text{in-vitro}} = 1.55 \frac{u_{\text{in-vitro}}}{u_{\text{in-vivo}}} Q_{\text{in-vivo}} \\
T_{\text{in-vitro}} = 1.55^2 \frac{u_{\text{in-vitro}}}{u_{\text{in-vivo}}} T_{\text{in-vivo}}
\]

Equation 1
Equation 2

Figure 4. In-vivo flow waves during natural breathing and with 30 l/min cannula flow (inspiration is positive), and the location of the 15 measurement times.

2.3 PIV measurements
The SPIV system consisted of a 15 Hz dual-head 120 mJ Nd:YAG laser (New Wave Solo XT), two digital 2 mega pixel CCD cameras (Dantec Flowsense) and optics to form a light sheet of approximately 2 mm thickness. The working liquid was seeded with near neutrally buoyant 10 µm hollow glass spheres and sequential images of the illuminated particles were recorded on 1600x1200 pixel frames at 10 Hz. The minimum pixel intensities of the following two particle images from the corresponding exposure were subtracted from each image to remove background noise. Non-flow regions were masked to impose a zero flow condition at the fluid-wall interface and to suppress wall reflections. Accurate mask images were created using a CT scan of the actual flow phantom and extracting cross-sections at the various measurement planes using the open source application Paraview (http://www.paraview.org/)

The particle images were divided into 80x80 pixels interrogation regions and the area average displacement calculated by locally cross-correlating the particle image intensities between two subsequent recordings. A grid spacing of 1.2 mm was used, giving an average overlapping factor of 75% and iterative window refinement was applied with a final interrogation window size was of 40x40 pixels. Window displacement and deformation based on the local velocity gradient was also applied. Displacement fields from 10 image pairs at corresponding time steps were ensemble averaged to yield phase averaged velocity fields.
The laser sheet and cameras were fixed and the reservoir and phantom traversed to take measurements at 30 sagittal slices through the nasal cavity at 1.5 mm increments. Image pairs were recorded at the 15 equi-spaced breath phases shown in Figure 4. Due to equipment and time constraints, the time delay between successive exposures was the same for each measurement phase. In order to maximise the resolution of slower phases the time delay was chosen such that the maximum particle displacement for the peak flow rate was 16 pixels. The time delay was chosen in this way for each measurement location traversed through the model. At a cross-section through the centre of the left nasal valve for example, the time delay was 1200 µs during natural breathing and 500 µs with cannula flow. Towards the outside of the inferior meatus where the flow velocities are slower, time delays in the order of 3500 µs and 2000 µs were used during natural and cannula breathing respectively.

The distance between the laser plane and the 8 mm thick, acrylic reservoir wall and therefore level of optical distortion was different for each traverse, requiring discrete camera calibrations for each measurement plane. Three calibration images were taken at 1 mm spacing per camera and traverse on a target plate that had 2.5 mm diameter white dots at 5 mm spacing (~1500 dots). Each dot was least squares mapped to a polynomial [7] to typically within an accuracy of 0.5 pixels. The cameras were separated by 43.8° and had an average magnification of 0.145 mm/pix.

### 3. Results and Discussions

All velocities shown are in-vivo scaled and only every third vector is displayed for clarity. The vector lengths denote in-plane velocity magnitude, and the colour contour shows absolute velocity calculated from all three components. Figures 5 and 6 show the velocity distribution through one sagittal cross-section of the nasal cavity that bisects the left nostril. Results are shown for 8 of the 15 time steps measured and their locations given in Figure 4. To enhance visualisation of lower velocities the velocity contour in Figure 6 has been capped at 8 m s\(^{-1}\).

During natural breathing, at the transition from inspiration to expiration where the net flow rate is zero, the maximum velocity in Figure 5.i is not zero but 0.4 m s\(^{-1}\). This velocity occurs in the nasal valve and is remnant of the jet of inspired air that enters the nasal cavity visible in Figures 5.vi–viii. This result clearly shows that the flow in the nasal cavity does have some time dependency. It can be seen in Figures 5.ii-iv that throughout expiration the velocity pattern is similar, with only the magnitude varying with the flow rate. A jet of air rises vertically from the constriction at location 8 in Figure 2 and attaches to the roof of the nasopharynx as it turns into the nasal cavity. Underneath this stream, on the nasopharynx floor there is counter clockwise vortex formed as the flow separates on the lee side of the bend. The main flow stream travels through the middle meatus before accelerating in the nasal valve and exiting the nostrils. Although the nasal valve has often been reported as the location of maximum velocities on expiration and inspiration, in this geometry on expiration the maximum velocity of 3.8 m s\(^{-1}\) occurs in the nasopharynx (Figure 5.iii).

On inspiration the flow accelerates in the nasal valve, reaching a maximum velocity of 3.3 m s\(^{-1}\) in Figure 5.vii at peak flow rate. At the rapid expansion from the nasal valve into the main cavity the flow separates, creating a stagnant region in the roof of the nasal cavity. Like on expiration the main flow stream passes through the middle meatus. Flow separation is also evident off the back of the inferior turbinate before the flow passes through the nasopharynx with a relatively even profile. Figures 5.vi and 5.viii correspond to an approximately equal inspiration flow rate, 5.vi occurring before and 5.viii occurring after the peak flow rate at position 5.vii. The velocity patterns in Figures
Figure 5. Phase averaged velocity fields through a cross section of the left nasal valve during natural breathing.
Figure 6. Phase averaged velocity fields through a cross section of the left nasal valve with 30 l/min cannula flow (no data available in the dark grey region)
5. vi and 5.viii are very similar, which would suggest time independence.

The flow pattern in the nasal cavity with NHF differed significantly from natural breathing, as visible in Figure 6. On expiration, high velocities were concentrated in the nasal valve, where cannula flow was forced to turn 180 degrees by the expired volume to additionally exit through the area available between the cannula’s prong and nostril. Two strong recirculating features were present above and below the jet. There was a more even flow distribution between the three meatus passageways with NHF. The momentum required to turn the jet, narrowness of the passageways, large velocities gradients creating high shear and low evenly distributed velocities upstream suggest there is a large pressure drop across the nasal valve with NHF. This resistance is thought to be largely responsible for the therapeutic levels of positive airway pressure experienced clinically. As with natural breathing the flow from the lungs entered the nasal cavity attached to the nasopharynx roof. On inspiration, a strong jet from the cannula rises towards and stays attached to the roof of the nasal cavity. The vortex above the jet persists throughout the entire respiratory cycle, however, on inspiration the vortex under the jet dissipates into a larger and weaker recirculatory feature. Maximal velocities of 16.5 ms\(^{-1}\) and 13.6 ms\(^{-1}\) on inspiration and expiration respectively, were both located in the cannula jet within the nasal valve. With NHF, timesteps of similar flow rate but different phase such as Figures 6.f and 6.g were nearly indistinguishable. These preliminary findings suggest that a quasi-steady assumption is valid for flow in the nasal cavity with NHF.

4. Conclusions

Time resolved SPIV has been used to measure the distribution and velocity of the airflow in the nasal cavity with physiologically reproduced pulsatile flows for natural breathing and breathing assisted with NHF. In-vivo flow measurements were Reynolds and Womersley number matched and applied in-vitro. An anatomically accurate transparent silicone flow phantom was created from CT images. Velocity maps were measured at 15 breath phases and at 30 sagittal slices traversed through the complete nasal cavity at 1.5 mm increments. The flow pattern in the nasal cavity differed significantly with and without NHF. Results have been shown that suggest a quasi-steady flow assumption for natural breathing is inaccurate, however, time dependency was not found with NHF.

5. References