Respiratory Physiology & Neurobiology 219 (2015) 51–57

Contents lists available at ScienceDirect

Respiratory Physiology & Neurobiology

journal homepage: www.elsevier.com/locate/resphysiol

Modeling the pharyngeal pressure during adult nasal high flow therapy

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ARTICLE INFO

Article history:
Received 9 February 2015
Received in revised form 22 June 2015
Accepted 22 June 2015
Available online 18 July 2015

ABSTRACT

Subjects receiving nasal high flow (NHF) via wide-bore nasal cannula may experience different levels of positive pressure depending on the individual response to NHF. In this study, airflow in the nasal airway during NHF-assisted breathing is simulated and nasopharyngeal airway pressure numerically computed, to determine whether the relationship between NHF and pressure can be described by a simple equation. Two geometric models are used for analysis. In the first, 3D airway geometry is reconstructed from computed tomography images of an adult nasal airway. For the second, a simplified geometric model is derived that has the same cross-sectional area as the complex model, but is more readily amenable to analysis. Peak airway pressure is correlated as a function of nasal valve area, nostril area and cannula flow rate, for NHF rates of 20, 40 and 60 L/min. Results show that airway pressure is related by a power law to NHF rate, valve area, and nostril area.

1. Introduction

Nasal High Flow (NHF) is a non-invasive ventilation therapy that provides respiratory support through the delivery of medical gas via wide bore nasal cannula. The therapy is postulated to provide benefit by promoting slow deep breathing or by flushing of CO_2 from the nasal cavity dead space (Dysart et al., 2009). The NHF offering from Fisher & Paykel Healthcare Ltd. (Auckland, New Zealand), Optiflow™ provides flows of heated and humidified gas at rates upto to 60 L/min, and delivers a low-level flow-dependent positive airway pressure (Dysart et al., 2009; Moller et al., 2015; Mundel et al., 2013). While NHF systems such as Optiflow™ – which delivers heated and humidified gas – are clinically attractive due to improved patient comfort and compliance (Maggiore et al., 2014; Roca et al., 2010), the lack of control of the delivered airway pressure has been noted as a clinical concern.

Previous studies have measured airway pressure and velocity at the oropharynx, nasopharynx, and via tracheostomy during NHF in patients and healthy volunteers (Dysart et al., 2009; Franke et al., 2014; Gerald et al., 2013; Groves and Tobin, 2007; Parke et al., 2009; Parke and McGuinness, 2013; Spence et al., 2012, 2011). There is considerable variation in measured airway pressure in these studies, for example mean expiratory pressure of 2.7 ± 1.0 cm H_2O in post-cardiac surgery patients for NHF at 35 L/min (Parke et al., 2009), and mean of 7.4 cm H_2O (range 5.5–8.8 cm H_2O) for NHF at 60 L/min in healthy volunteers (Groves and Tobin, 2007). The NHF system OptiFlow™ is typically used with cannula flow rates between 20 and 60 L/min in the adult population. In a clinical setting, if a patient is not responding satisfactorily to NHF, the clinician may choose to increase NHF rate, resulting in an increase of positive end-expiratory pressure (PEEP). Informing the process of titrating an efficacious delivered pressure with NHF is clinically very significant in terms of effect on both cardiovascular and lung function.

The geometry of the nasal airways in the adult population varies widely, although with some consistent trends (Liu et al., 2009; Taylor et al., 2010). For example, the nasal valve—is a constriction located posterior to the nasal vestibule and generally represents the minimum cross-sectional area of the nasal airway. The nasal valve may generally vary in range of 40–120 mm^2 (Liu et al., 2009; Taylor et al., 2010) and the nostril area between 80 and 150 mm^2. Groves and Tobin (2007) observed a linear increase in expiratory pressure with flow, and they postulated that the facial features, particularly the smaller nares of females, could explain the higher pressure observed in some groups (Parke and McGuinness, 2013). Hence it is plausible that a relationship between airway pressure, nasal airway geometry, and cannula flow can be quantified.

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http://dx.doi.org/10.1016/j.resp.2015.06.011
1569-9048/

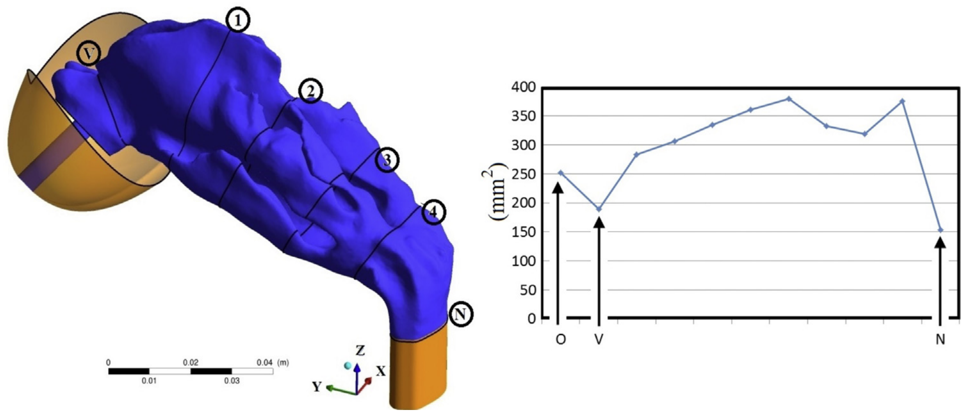

Fig. 1. Reconstructed 3D geometry of an adult nasal airway. (a) Lateral view. (b) Cross-sectional area along downstream locations. (c) Cross-sectional shapes within the nasal passage. 'V' is nasal valve. 'O' is nostril. 'N' is nasopharynx.

Computational fluid dynamics (CFD) uses numerical methods and algorithms to run computer simulations of fluid flows and is now frequently used to simulate airflow in nasal or bronchial airways (Bates et al., 2015; Doorly et al., 2008a,b; Elad et al., 2008; Quadrio et al., 2014; Schroeter et al., 2014; Subramaniam et al., 1998; Tawhai and Lin, 2010; Taylor et al., 2010). In the current study we have used CFD to elucidate the producing mechanism and maximum expected magnitudes of nasopharyngeal pressures during supply of NHF. Maximum steady-state inspiratory and expiratory nasopharyngeal pressures during supply of NHF are estimated during mouth closed breathing. Although in clinical practice a patient receiving NHF may breathe with their mouth open, highest airway pressures are generated with the mouth closed (Parke et al., 2011). Flow trajectories and static pressure in an anatomically structured 3D model of a single nasal airway are considered in detail. The construction of anatomically-structured geometry to simulate pharyngeal pressures across patient populations is challenging due to limited access to imaging (when imaging is available) and the time involved in image processing, 3D model construction, and CFD simulation. To address this issue, a 'minimal' airway is presented that captures the important geometric properties of the anatomical airway. Previous studies have adopted a similar approach to estimate nasal resistance (Javaheri et al., 2013; Schreck et al., 1993). The minimal airway geometry is derived by simplifying the cross-sectional shapes of the detailed anatomical airway. The geometry is then manipulated to investigate the effect of subject variability in nasal valve area and external nostril area on static pressure drop along the airway. Investigating variability has potential clinical relevance for efficacy across the differing patient demographic.

2. Methods

2.1. Anatomical model geometry

Volumetric computed tomography (CT) imaging was used to obtain a three-dimensional (3D) geometry of a human nasal airway. The reconstructed 3D geometry is shown in Fig. 1. The cross-sectional area along the model centerline is plotted in Fig. 1(b) and cross-sectional shapes along the nasal passage are shown in Fig. 1(c). The airway has nostril areas of 119 mm^2 and 133 mm^2 for left and right nostril, respectively, and nasal valve areas of 94 mm^2 (left) and 96 mm^2 (right).

Since closed-mouth breathing was assumed in this study the oral cavity was not included. With closed mouth breathing the highest pressure scenario is therefore simulated. The nasal air was assumed to have the same density and dynamic viscosity as the 37 °C 100% relative humidity air delivered by the cannula's nasal prongs into each nostril. The cannula geometry was based on Fisher & Paykel Healthcare's medium sized adult Optiflow™ cannula (OPT844) that has an elliptical prong cross-section (major axis 5.5 mm and minor axis 3.4 mm). The presence of the nasal cannula prong reduced the nostril space to 104 mm^2 (left) and 118 mm^2 (right), respectively.

2.2. Minimal model geometry

A 3D minimal airway model was developed for one half of the nasal airway, using the cross-sectional area at 11 locations along the anatomical nasal passage. At these locations, the complex nasal airway cross-sectional shape was replaced with simple shapes as shown in Fig. 2, while maintaining the cross-sectional area at these relative locations. The nasal airway geometry was created in SolidWorks™ using interpolation and B-spline surface extrusion. To study the impact of geometry differences, nasal valve areas of 98, 70 and 50 mm^2, and nostril areas of 120, 90 and 72 mm^2 were selected while cross-sectional areas in other locations were not changed. The centerline cross-sectional areas for different combinations are given in Fig. 2(b). Each combination was assigned a geometry number: 11, 12, 13, 22, 23 and 32. For example, geometry 11 has nostril area of 120 mm^2 and valve size of 98 mm^2 and so on. Details of these geometries are given in Table 2.

2.3. Numerical methods

Steady-state inspiratory and expiratory flows were simulated using ANSYS CFX 14.0 (ANSYS, Inc.), a fluid-dynamics solution package. The tracheal flow rate and NHF (or cannula flow) rate are inputs to the simulation. NHF has been reported to reduce respiratory rate (Rittayamai et al., 2014; Roca et al., 2010; Sztrymf et al., 2011), increase tidal volume (Corley et al., 2011; Mundel et al., 2013) and reduce minute ventilation (Braunlich et al., 2013). In practice, meeting or exceeding peak inspiratory volumes ensures accuracy of delivered oxygen concentrations when there is no requirement to entrain air to meet demand (Ritchie et al., 2011). Steady flowrates representing peak inspiration and peak expiration of 36 L/min and 20 L/min, respectively, were assumed (Spence, 2011). Neglecting flow bias these flows corresponded to 18 L/min and 10 L/min through each nostril, respectively. The flow rates do not correspond to a particular minute volume, as only steady-state flows were considered and not the time period over which the flow would occur. Nevertheless, the flowrates of Spence et al. were taken on a healthy 23-year old male (height 184 cm, weight 85 kg) with a resting minute volume of 7.0 L/min. Three NHF rates (Q_C = 20, 40 and 60 L/min) were considered in our study. A NHF-Reynolds num-