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# Model identifies causes of nasal drying during pressurised breathing

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ABSTRACT

Patients nasally breathing pressurised air frequently experience symptoms suggestive of upper airway drying. While supplementary humidification is often used for symptom relief, the cause(s) of nasal drying symptoms remains speculative. Recent investigations have found augmented air pressure affects airway surface liquid (ASL) supply and inter-nasal airflow apportionment. However the influence these two factors have on ASL hydration is unknown.

The purpose of this study is to determine how ASL supply and airflow apportionment affect ASL hydration status for both ambient and pressurised air breathing conditions. This is done by modifying and adapting a nasal air-conditioning and ASL supply model.

Model predictions of change in inter-nasal airflow apportionment closely follow *in-vivo* results and demonstrate for the first time abnormal ASL dehydration occurring during augmented pressure breathing.

This work quantitatively establishes why patients nasal breathing pressurised air frequently report adverse airway drying symptoms. The findings from this investigation demonstrate that both nasal airways simultaneously experience severe ASL dehydration during pressurised breathing.

## 1. Introduction

Positive air pressure therapy users frequently report upper airway drying symptoms. Up to 65% of obstructive sleep apnoea syndrome (OSAS) patients receiving nasal applied continuous positive air pressure therapy (*n*-CPAP) report nasal complications suggestive of mucosal drying (Mador et al., 2005; Malik and Kenyon, 2004). The most prevalent complaints include nasal dryness, crusting, congestion, sneezing, rhinorrhoea and itching (Arfoosh and Rowley, 2008; Kalan et al., 1999; Kreivi et al., 2010; Massie et al., 1999). Nasal drying symptoms during *n*-CPAP therapy have been attributed to unidirectional airflow created by mouth leaks (Worsnop et al., 2009). However other work has shown these symptoms are also common when mouth leaks are absent (Kalan et al., 1999), and that other factors may contribute to mucosal drying (Constantinidis et al., 2000; Devouassoux et al., 2007; Malik and Kenyon, 2004; Worsnop et al., 2009). The use of supplementary humidification is the most common method used to treat dryness symptoms and improve patient comfort (Arfoosh and Rowley, 2008; Koutsourelakis et al., 2011; Malik and Kenyon, 2004), but the causes(s) of nasal airway drying during *n*-CPAP breathing currently remain unknown.

### 1.1. Nasal functionality

The human nose serves an important role in maintaining airway health by heating and humidifying inhaled air as well as entrapping inhaled pathogens and pollutants (Keck et al., 2000). To achieve these functions the airway is lined with an airway surface liquid (ASL), which acts as a reservoir of heat and water supplied to the air from the underlying mucosa (Warren et al., 2010). This storage provides a thermal and water buffer between cyclic breathing demands and the more constant cellular and vascular supplies. The ASL also plays a crucial role in mucociliary transport (Boek et al., 2002), and as such the height of this layer must be maintained within tight limits (Williams et al., 1996). ASL is provided by mucosal glandular and cellular sources. Cellular ASL supply has been shown to be stimulated by pressure induced tidal breathing stresses (Button and Boucher, 2008). Previous investigations have shown the ASL purinergic supply is drops by 22% for the pressure range spanning 5–15 cm H2O pressure (White et al., 2014) so this remains fixed at this reduced value. The influence pressure elicited reduction in ASL cellular supply has on nasal ASL hydration during *n*-CPAP breathing is currently unknown.

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### 1.2. Inter-nasal airflow apportionment (nasal cycle)

Normal nasal airflow alternates in dominance between the two nostrils with an ultradian rhythm called the "nasal cycle," which occurs approximately every 90 min during waking and every 180 min during sleep (Rohrmeier et al., 2014). Computational modelling has quantitatively demonstrated that the nasal cycle enables the upper airway to accommodate the contrasting roles of air humidifying and warming, and the removal of entrapped contaminants through fluctuation in airflow partitioning between each airway (White et al., 2015). The airway passing more air is termed 'patent', while the other is known as 'congested'. A recent in-vivo study has shown that normal airflow apportionment between the 'patent' and 'congested' airways in healthy individuals (n = 20) is abolished during *n*-CPAP breathing (White et al., 2016), resulting in the previously 'congested' airway passing a greater apportionment of the airflow, while the formerly 'patent' airway conducts a lesser amount. Unlike ASL water supply, only airway geometry has previously been shown to vary over a range of CPAP pressures.

It is the intention of the work reported in this paper to investigate the significance that pressure elicited changes in ASL water supply and inter-nasal airflow apportionment have on ASL drying during simulated *n*-CPAP breathing. It is noted that this work will investigate the behaviour of the system without the nasal cycle, i.e.: while the passageways are in a relatively stable airflow apportionment state for a specific individual's nasal airway.

## 2. Method

In order to investigate ASL drying as a function of ambient and pressurised breathing, it was necessary to link data and models which have previously been used for other purposes. The core of the method consisted of the state-variable tidal breathing model used to predict inter-nasal apportionment flows and ASL hydration levels for ambient pressure breathing (White et al., 2015). This model had been developed using nasal passageway geometry and ASL fluid supply data, previously for ambient (sea level) pressure conditions. This model was adapted to deal with pressurised breathing using two modifications. Firstly, MRI derived data, which had been used to determine pressure elicited changes in inter-nasal geometry and airflow partitioning, was added (White et al., 2016). Secondly, a fixed value of maximal ASL fluid supply data under pressurised conditions was also included (White et al., 2014) as there is no evidence in the current literature that supports the view that ASL fluid supply is time dependent. This adapted model was then able to predict the airflow partition ratio (percent of total airflow through each nasal passage) and ASL hydration levels along each individual nasal passage for both ambient and augmented pressure breathing. With typical *n*-CPAP pressure augmentation ranging from 4 to 20 cm $H_2O$ pressure, we have chosen to test our model at a mid-range pressure of 10 cm $H_2O$. Simulated air conditions representative of comfortable indoor conditions with temperature of 23 °C and relative humidity of 45% were used.

### 2.1. Inter-nasal airflow apportionment

The model considers the nasal cavity as a parallel resistance network, each containing a series of aligned tubes of varying hydraulic diameter between each naris and the nasopharynx. Model predictions of nasal resistance which dictates inter-airflow apportionment between each airway are based on local air velocities and nasal geometry, derived from MRI scans (White et al., 2016). Summing the resistances of each discrete model lump along the left and right airways enables calculation of specific airflow resistance for each nasal passageway. Total airway resistance is found by considering both nasal airways as a parallel resistance network and calculation of the inter-nasal airflow partitioning ratio is based on ratios of total and individual airway airflow resistances. Pressure elicited change in inter-nasal airflow apportionment is predicted by applying a pressure-based correction factor to the nasal geometry along each airway. In the first part of this investigation, we compare the predicted inter-nasal airflow partitioning to measured values for one individual between ambient and augmented pressure breathing conditions. The actual results are derived from instantaneous inter-nasal airflow measurement from an individual during 20 min of ambient breathing before receiving pressurised air for a further 20 min.

### 2.2. ASL hydration

Change in ASL hydration status within the nose predicted by the model is determined from the difference between mucosal fluid supply and humidification fluid demand throughout a full breath cycle. The maximal ASL fluid supply used by the model has previously been estimated to be 7.9 g/cm$^2$-hr during simulated ambient air pressure breathing (White et al., 2015), however this has been shown to reduce by 22% during augmented air-pressure breathing above 5 cm $H_2O$ pressure (White et al., 2014). This revised fluid supply value is implemented within the model during simulated augmented pressure breathing. Hydration status is presented in terms of, ASL water equivalent height, ($H_{e,ASL}$), to account for differing water loss from each of the two liquid layers (White et al., 2015). Within the model we consider fully hydrated ASL to have an $H_{e,ASL}$ of 10 μm, and when severely dehydrated, $H_{e,ASL}$ = 0 μm. Mucociliary transport is deemed to have ceased when the ASL becomes severely dehydrated. Data collection protocols for previous studies were approved by the Auckland University of Technology Ethics Committee under ethics application numbers 14/02 (airflow study) and 10/121 (geometry study), and were conducted in accordance with the Declaration of Helsinki.

## 3. Results

### 3.1. Inter-nasal airflow apportionment

Model predictions of inter-nasal airflow apportionment for both nasal airways, shown in Fig. 1 for one individual, follow closely the

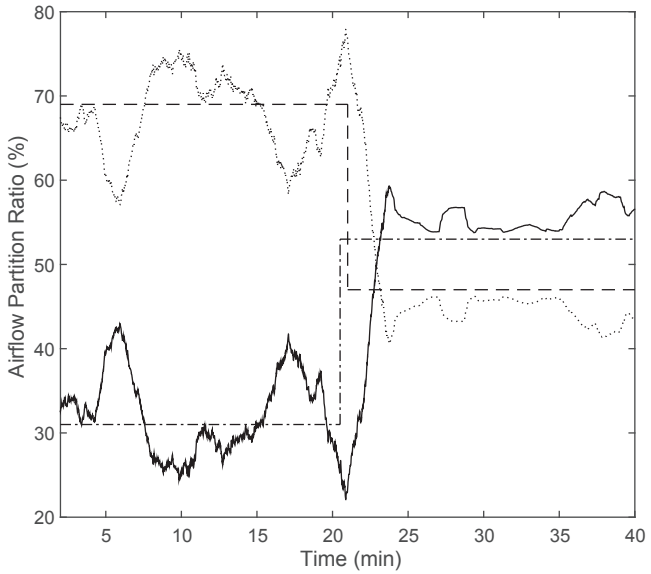


**Fig. 1.** Measured *in-vivo* and predicted inter-nasal airflow apportionment ratio during the first 21 min of ambient and a further 19 min during *n*-CPAP breathing at 10 cm $H_2O$ pressure. Predicted model airflow partition ratios are shown as straight lines as these are based on single MRI image data while measured values fluctuate over time.⋯⋯ = measured right airway, ____ = Measured left airway, −·−·−·− = predicted left airway, _ _ _ = predicted right airway.