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Glottal Motion and its Impact on the Respiratory Flow

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1 Introduction
Advantages of inhaled therapies as a priori targeted supply of drugs make them particularly convenient for the treatment of lung diseases. Nevertheless, several physical and anatomical factors can largely influence treatment efficiency. In particular, upper airways (UA) anatomic arrangement acts as an unwanted filter, which limits the amount of drug delivered to the lung [5]. More specifically, the glottis, defined by vocal folds aperture within the larynx, causes airways to narrow in a minimal transition cross-section. This anatomical constriction with a triangular dynamic mesh of 24000 elements. A pressure outlet boundary condition was set to 0 Pa. Unsteady boundary conditions comprising the glottal width d(t) and the velocity inlet were parametrically varied in agreement with the in vivo study. A no-slip shear boundary condition was applied at solid walls. Initially, zero velocities and pressures were assumed at all points. Equations were solved using a first-order time and spatial discretization schemes, and a time step set to 0.12s.

2 Methods

In vivo study. In vivo experiments were conducted in the ENT department of La Timone Hospital. One healthy female volunteer (LB, age 29) and one healthy male volunteer (OB, age 48) were recorded while performing two 30s breathing tasks: normal breathing ( eupnea) and forced breathing ( tachypnea). Laryngofiberscopic investigations were made using a flexible nasofiberscope (Storz 202220 20 tricam camera) with a continuous cold light source and a color CCD camera. Laryngeal images were captured with a camera frame rate of 25 frames/s and an image resolution of 768x288 pixels. The oral airflow signal was simultaneously registered by means of a pneumotachograph placed at the mouth, EVA2 [7].

In silico study. As a first approximation of the glottal geometry was built a 2D rectangular moving constriction with a triangular dynamic mesh of 24000 elements. The mesh density ensures grid-independent results. CFD simulations were conducted using Fluent 6.3.26 under laminar airflow conditions, assuming an incompressible Newtonian gas of viscosity \( v \) equal to \( 1.789 \times 10^{-5} \) kg/ms. A pressure outlet boundary condition was set to 0 Pa. Unsteady boundary conditions comprising the glottal width \( d(t) \) and the velocity inlet were parametrically varied in agreement with the in vivo study. A no-slip shear boundary condition was applied at solid walls. Initially, zero velocities and pressures were assumed at all points. Equations were solved using a first-order time and spatial discretization schemes, and a time step set to 0.12s.

3 Results and discussion
Airflow rate. In vivo measurements yielded to about 30 respiratory cycles during eupnea and 40 during tachypnea. Every respiratory cycle was detected using a zero-tracking method developed in Matlab R2011b. Each airflow signal \( Q \) was normalized with respect to the maximum value achieved within the cycle, \( Q_{\max} \). Time \( t \) was normalized by corresponding respiratory period, \( T = 2\pi/\omega \). The maximal period registered within each 30s sequence is noted \( T_{\max} \). Signals were finally averaged into one mean flow-rate, herein noted \( \langle Q/Q_{\max}\rangle \). Figure 1a illustrates the typical flow-rate \( \langle Q/Q_{\max}\rangle \) as a function of \( t \), produced by subject OB during eupnea (maximal \( Q_{\max} = 1.9 \) dm/s, \( T_{\max} = 0.62s \)) and tachypnea (maximal \( Q_{\max} = 4.3 \) dm/s, \( T_{\max} = 4.8s \)). In comparison, a sinusoidal evolution is plotted. Conventionally, positive (resp. negative) flow-rate values correspond to expiration (resp. inspiration) phase. During eupnea, the inspiration and expiration curves are similar (up to sign) and their durations are approximately equal, as commonly found in the literature [6]. During tachypnea, the mean flow-rate curve deviates from the harmonic signal although inspiration and expiration durations remain roughly equal. A phase difference of about 22° in flow-rate maximal occurrences has been measured between sinusoid and eupnea curve, which corresponds to 6% of the breathing period. The phase difference increases during tachypnea up to 58°, namely 16% of the cycle duration. LB breathing shows similar characteristics.

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Glottal motion. Glottal motion was extracted from the laryngoscopic images using Matlab Image Processing Toolbox and different phases: (i) Focus on a region of interest using a cross-correlation technique, (ii) Smoothing using a specific filter function [10], (iii) Detection of the glottal contours using a segmentation method [2] (iv) Measurement of the glottal antero-posterior diameter $AP_g$, glottal area $A_g$ and glottal width $d_g$ (see Fig. 1b), (v) Normalization by comparing $AP_g$ diameter with the initial value $AP_g^0$, assumed as a geometrical invariant [8], (vi) Conversion from pixels to millimeters, assuming $AP_g = 22$mm [5].

Figure 1: (a) Mean flowrate $\langle Q/Q_{max}\rangle$ measured during cycles of eupnea and tachypnea (subject OB) and comparison with sinusoid. (b) Illustration of glottal image post-processing. (c) Detected glottal area $A_g$ and glottic width $d_g$ and flowrate $Q$.

![Figure 1](image1.png)

Figure 2: 2D unsteady simulations of velocity field through the moving glottis during a breathing cycle.

4 Conclusions

The in vivo study showed that the glottis can be extremely variable during breathing and hence influence airflow characteristics. A glottal area widening was quantified during inspiration, with a typical ratio of 3:1 as compared to expiration. Airflow rate variations differ from harmonic signal during eupnea as well as tachypnea. The correlation between flow-rate and glottal area will be discussed and compared to previous clinical investigations. Preliminary 2D CFD simulations of the glottal jet were performed, based on the measured flow-rate and glottal changes during eupnea. Impact of unsteady flow conditions on the jet development is demonstrated.

References