Realistic glottal motion and airflow rate during human breathing

Adam Scheinherr, Lucie Bailly, Olivier Boiron, Aude Lagier, Thierry Legot, Marine Pichelin, Georges Caillibotte, Antoine Giovanni

Abstract

The glottal geometry is a key factor in the aerosol delivery efficiency for treatment of lung diseases. However, while glottal vibrations were extensively studied during human phonation, the realistic glottal motion during breathing is poorly understood. Therefore, most current studies assume an idealized steady glottis in the context of respiratory dynamics, and thus neglect the flow unsteadiness related to this motion. This is particularly important to assess the aerosol transport mechanisms in upper airways.

This article presents a clinical study conducted on 20 volunteers, to examine the realistic glottal motion during several breathing tasks. Nasofibroscopy was used to investigate the glottal geometrical variations simultaneously with accurate airflow rate measurements. In total, 144 breathing sequences of 30 s were recorded.

Regarding the whole database, two cases of glottal time-variations were found: “static” or “dynamic” ones. Typically, the peak value of glottal area during slow breathing narrowed from 217 ± 54 mm² (mean ± STD) during inspiration, to 178 ± 35 mm² during expiration. Considering flow unsteadiness, it is shown that the harmonic approximation of the airflow rate undervalues the inertial effects as compared to realistic patterns, especially at the onset of the breathing cycle. These measurements provide input data to conduct realistic numerical simulations of laryngeal airflow and particle deposition.

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1. INTRODUCTION

Inhaled therapies play a major role in the treatment of lung diseases like asthma or chronic obstructive pulmonary diseases. Basic advantage of aerosol therapy lies in the direct delivery of high local concentrations of the drug to the site of action [1]. However, characteristics of inhaled particles, airways morphology, carrier gas and flow properties can largely influence the transport mechanisms and treatment efficiency [2–5]. Particularly, the upper airways (UA) anatomic arrangement can act as an unwanted filter, which limits the amount of drug delivered to the lungs. Recent clinical experiments have been conducted to quantify the distribution of radiolabeled aerosols in the human airways using combined single photon emission computed tomography (SPECT) and X-ray computer tomography (CT) [6,7]. It was shown that particle deposition in extra-thoracic region can reach as much as 40% of the inhaled mass in the worst cases.

More specifically, within the larynx, the glottis (the space between vocal folds) narrows the airways to a minimal transition cross-section. Therefore its geometrical variations can affect breathing flow resistance [8–11]. This anatomical singularity yields to a complex jet-like glottal airflow, important recirculation zones and a locally turbulent behavior [12–19], which can be determinant on particle deposition by inertial impaction [3,20,21]. Regarding the glottal geometry’s impact on the tracheal flow during breathing, a devoted description has been given by Brouns et al. [18,19]. This numerical study demonstrates the effect of the glottal size and shape on the overall fluid dynamics behavior, using a 3D idealized model of upper airways comprising a static glottis of parametrical aperture. This typical flow alteration can yield to the rise of the tracheal deposition of nano- and micrometer particles by factors ranging from 2 to 10 [20]. However, as pointed out in Brouns et al. [18], there is a current need for rendering the design of future upper airway models even more realistic, including glottal shape, area and kinetics in correlation with a given inlet inhalation flow rate. Thus, the knowledge of realistic
glottal motion during human breathing will enable to further assess the filtration efficiency of the upper airways.

In this context, the purpose of the present study is to determine the glottal motion during several oral breathing tasks, and to investigate the correlation between this motion and the measured breathing flow rates. The first in vivo observation of the glottal motion dates back to the 19th century, with Garcia’s laryngeal mirror [22]. Since the 1980s, advances in the medical equipment have allowed a refined exploration of the vocal-fold dynamics using laryngoscopy [9,10], high-speed cinematography [23,24], videokymography [25], electroglottography [26] or photoglottography [27]. These experimental techniques were extensively used to characterize the vocal-fold vibrations during human phonation (e.g. see a review by Ziethe et al. [28]). By contrast, however, the glottal variations during different human breathing regimes have been barely investigated so far.

Despite a few reference studies [9,10,29], the relationship between glottal motion and breathing capacity in specific respiratory cases of controlled breathing, for which the subjects were asked to synchronize their breathing frequency with a metronome projected on the respiration, tasks (i) and (ii) were acquired in condition of spontaneous breathing. By contrast, the other tasks corresponded to cases of controlled breathing, for which the subjects were asked to synchronize their breathing frequency with a metronome projected on an instruction computer. These tasks intended to explore the extent of glottal motion and breathing capacity in specific respiratory contexts. In the end, 18 volunteers only (9 females and 9 males) successfully performed the entire protocol, thus yielding to a database comprising 144 sequences of 30s.

2.1.3. Measurements

All measurements were done in the seated posture. The glottis was observed using a flexible nasofiberscope equipped with a PAL camera (Storz endovision XL 202800) and a continuous cold light source. Laryngeal images were captured at a frequency of 25 Hz (768 × 288 pixels). The airflow rate was simultaneously registered by means of a pneumotachograph placed at the mouth, the EVA2™ system (S.Q.Lab, www.sqlab.fr) [30]. It consists of a two-grid flowmeter characterized by a small dead volume, specific linearized response detected on the airflow signal, and an accuracy of 1 cm³s⁻¹. The sampling frequency of the flow rate signal was 6250 Hz. A trigger generated by an acquisition tool developed in NI LabWindows™/CVI was used to synchronize the recordings of laryngeal images and flow rate signal. Note that for several subjects, a local anesthetic (Lidocaine Aiguettant 5%) was sprayed in the naris before the fiberscope introduction, so as to provide a better comfort during the invasive examination.

Ambient temperature T_a [K] was also measured.

Within each recorded 30s-sequence, every respiratory cycle was detected on the airflow signal, Q, using a zero-tracking method. Conventionally, positive and negative flow rate values correspond to inspiration and expiration, respectively. A BTPS correction (Body Temperature Pressure Saturated) was applied to convert the flow measured at ambient conditions to the thermodynamic conditions measured at ambient conditions to the thermodynamic conditions.

2.2. Data processing

All data was processed using Matlab®. Any point in the upper airways was located by the (x, y, z) coordinates as introduced in Fig. 1.

2.2.1. Airflow rate

Within each recorded 30s-sequence, every respiratory cycle was detected on the airflow signal, Q, using a zero-tracking method. Conventionally, positive and negative flow rate values correspond to inspiration and expiration, respectively. A BTPS correction (Body Temperature Pressure Saturated) was applied to convert the flow measured at ambient conditions to the thermodynamic conditions.
Corresponding contour of glottal area \(A_g\), antero-posterior diameter \(A_P\) and glottal width \(d_g\) at the time of maximum opening \(t_1\) (c) and at time of minimum opening \(t_2\) (d).

2.2.2. Glottal motion detection

This procedure was limited to the analysis of two major tasks, \(\text{Eup}_{15}\) and \(\text{Tachyp}_{30}\), altogether representing slow and rapid breathing tasks respectively. At first, breathing cycles, with hidden glottis aperture (by epiglottis or arytenoid cartilages) were removed from the video processing. Finally, about 126 (252) respiratory cycles out of 149 (291) were associated to task \(\text{Eup}_{15}\) (\(\text{Tachyp}_{30}\)). For these retained sequences, the glottal motion was extracted from each laryngoscopic image \(k\) taken at shot-instant \(t_k\) as follows: \(i\) correction of \(x\)- and \(y\)-motions of the fibrescope and focus on a region of interest using a customized cross-correlation technique; \(ii\) smoothing of the resized image using a Gaussian bandpass filtering; \(iii\) detection of the glottal area \(A_g(t_k)\) contours applying a geodesic active contour-based method [32], as shown in Fig. 1c; \(iv\) determination of antero-posterior diameter \(A_P(t_k)\) (see Fig. 1d); at first 2 points were detected on left and right vocal-fold contour to identify the left and right glottal border lines; then, from the intersection of those border lines was derived a centerline; at last \(A_P\) was defined by the 2 cross points of the centerline with the vocal-fold contour; \(v\) determination of glottal width \(d_g(t_k)\) (see Fig. 1d), as a perpendicular line to \(A_P\) at its mid-level; \(vi\) correction of vertical movement of the larynx and \(z\)-motion of the fibrescope by assuming \(A_P\) as a geometrical invariant during breathing [33]; \(vii\) finally, conversion of the measured data from pixels to millimeters, assuming \(A_P = 22.1\) mm for males and \(17.6\) mm for females [34]. Note that the relevance of these latter values was preliminary checked using other means of clinical investigation allowing an easier spatial calibration, such as the high resolution computed tomographic scans recorded by Fleming et al. [6]. In particular, a typical value of 22 mm was confirmed by this technique in the case of a male adult.

2.2.3. Methodology accuracy

The measurement of the distance between the camera extremitiy and the glottal plane would enable direct dimensions conversion from pixels to millimeters. However, touching the glottal plane was not allowed as it could cause laryngospasm. Therefore, we chose the hypothesis of \(A_P\) being a geometrical invariant [33] as an alternative to convert our pixels measurements. This assumption was previously applied by several authors [10,35,36]. The accuracy of the protocol was tested by recording a printed glottal image while simulating motions of the camera along \(x\), \(y\)- and \(z\)-directions. The glottal dimensions were detected with an average error smaller than 5%. Nevertheless, depending on the recorded laryngoscopic image quality (e.g. variable light contrast), the maximal error reached by the detection procedure was assessed at 10%.

3. RESULTS

In the followings, if \(X\) is a function of time \(t\), \(X_{\text{max}}\) refers to the maximum value of \(X\) over duration \(t\).

3.1. Airflow-rate pattern

For each task and gender of the database, Table 2 summarizes the measured primary respiratory variables \(T_I\) [s], mean inspiration airflow rate peak \(Q_{\text{max}}\) [L.min\(^{-1}\)] and inspired volume \(V_I\) [L]. Derived ratios \(T_I/T_{\text{p}}\) (\(T_{\text{p}}\) [s] being the mean inspiration period, see Fig. 1a), \(Q_{\text{max}}/Q_{\text{max}}\) (\(Q_{\text{max}}\) [L.min\(^{-1}\)] being the mean expiration airflow rate peak), and the BTPS correction factor \(C_E\) are also reported, as well as the number of processed breathing cycles \(N^e\). Figs. 2 and 3 present the corresponding normalized mean flow-rates \(Q/Q_{\text{max}}\) as a function of \(\omega\) [rad], where \(\omega\) is the pulsation of the average cycle defined as \(\omega = 2\pi / T_I\). The flow-rates obtained for the slow breathing tasks (i) to
Table 2
Measured airflow rate parameters in function of gender and breathing task (see chapter Airflow rate pattern in Results). Gray lines are the tasks further treated for the glottal motion analysis.

<table>
<thead>
<tr>
<th>Task</th>
<th>Eupf</th>
<th>Eup15</th>
<th>Hyperf</th>
<th>Aerosol</th>
</tr>
</thead>
<tbody>
<tr>
<td>Males</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Eupf</td>
<td>2.96</td>
<td>1.5–5.8</td>
<td>2.10</td>
<td>1.43–3.45</td>
</tr>
<tr>
<td>Eup15</td>
<td>3.99</td>
<td>3.7–4.5</td>
<td>2.26</td>
<td>1.82–3.04</td>
</tr>
<tr>
<td>Tachyp30</td>
<td>2.00</td>
<td>1.8–2.4</td>
<td>2.13</td>
<td>1.69–2.55</td>
</tr>
<tr>
<td>Tachyp60</td>
<td>1.02</td>
<td>0.8–1.3</td>
<td>2.10</td>
<td>1.72–2.99</td>
</tr>
<tr>
<td>Tachyp90</td>
<td>0.75</td>
<td>0.6–0.9</td>
<td>2.11</td>
<td>1.70–3.02</td>
</tr>
<tr>
<td>Hyperf</td>
<td>7.39</td>
<td>3.5–12</td>
<td>2.20</td>
<td>1.73–2.67</td>
</tr>
<tr>
<td>Aerosol</td>
<td>3.29</td>
<td>1.4–5.8</td>
<td>3.55</td>
<td>2.09–6.82</td>
</tr>
<tr>
<td>Females</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Eupf</td>
<td>4.00</td>
<td>2.2–3.5</td>
<td>2.36</td>
<td>1.56–2.36</td>
</tr>
<tr>
<td>Eup15</td>
<td>3.99</td>
<td>3.6–4.1</td>
<td>2.21</td>
<td>1.94–2.15</td>
</tr>
<tr>
<td>Tachyp30</td>
<td>1.99</td>
<td>1.8–2.1</td>
<td>2.12</td>
<td>1.81–2.55</td>
</tr>
<tr>
<td>Tachyp60</td>
<td>1.00</td>
<td>0.8–1.2</td>
<td>2.10</td>
<td>1.59–2.84</td>
</tr>
<tr>
<td>Tachyp90</td>
<td>0.73</td>
<td>0.4–1.1</td>
<td>2.10</td>
<td>1.63–3.55</td>
</tr>
<tr>
<td>Hyperf</td>
<td>8.00</td>
<td>5.1–13.4</td>
<td>2.72</td>
<td>1.80–2.77</td>
</tr>
<tr>
<td>Aerosol</td>
<td>3.19</td>
<td>1.9–5.5</td>
<td>4.74</td>
<td>2.49–13.04</td>
</tr>
</tbody>
</table>

Fig. 2. Normalized mean flow rates $\dot{Q}/\dot{Q}_{\text{max}}$ as function of $\omega t$ [rad] for slow breathing tasks (Eupf, Eup15, Hyperf) (Panels a and c for males and females, respectively) and rapid breathing tasks (Tachyp30, Tachyp60, Tachyp90) (Panels b and d for males and females, respectively).
(iv) (see Figs. 2a and 2c for males and females respectively), and rapid breathing tasks (v) to (vii) (see Figs. 2b and 2d) are displayed. The flow-rate obtained for the specific inhalation task (viii) is plotted in Fig. 3.

3.1.3. Breathing amplitude

Values of the flow-rate maximal amplitudes $\frac{Q_{\text{max}}}{Q_{\text{mean}}}$ are very different inter-tasks and inter-subjects, as detailed in Table 2. Over the whole database, the amplitudes are 30% higher for male than for female subjects. During spontaneous breathing, typical peak values ranged between 29 L.min$^{-1}$ and 68 L.min$^{-1}$, with a mean value of 43 L.min$^{-1}$ for males, and 32 L.min$^{-1}$ for females. These results are in line with previous studies (see range of values from 13 L.min$^{-1}$ to 78 L.min$^{-1}$ in Proctor & Hardy [38]).

3.2. Glottal motion

Table 3 summarizes the primary glottal dimensional parameters ($A_{\text{g}}$, $A_{\text{g}}^{\text{max}}$, and $A_{\text{g}}^{\text{min}}$) measured during slow breathing (Eup$\text{p}_{15}$) and rapid breathing (Tachyp$\text{p}_{30}$). Area $A_{\text{g}}$ is the mean glottal area obtained over the average cycle $\overline{Q}$; $A_{\text{g}}^{\text{max}}$ is the mean glottal area obtained during inspiration (resp. $A_{\text{g}}^{\text{max}}$ during expiration), and $A_{\text{g}}^{\text{min}}$ is the peak value of $A_{\text{g}}^{\text{max}}$ area (resp. $A_{\text{g}}^{\text{min}}$ is the minimum value of $A_{\text{g}}^{\text{max}}$). The derived ratios $A_{\text{g}}^{\text{max}}/A_{\text{g}}^{\text{min}}$, representing the ratio of the mean (resp. peak) $A_{\text{g}}$ amplitude during inspiratory and expiratory phases, are also reported. The variations of $A_{\text{g}}^{\text{max}}$ in function of $\overline{Q}$ for all subjects are shown in Figs. 4 and 5, for slow and rapid breathing respectively.
Table 3

Measured glottal dimensions parameters in function of gender, group of subjects and breathing rate. Gray lines are the groups for which a time-varying glottal area was detected during the average breathing cycle.

<table>
<thead>
<tr>
<th>Slow Breathing–Eupnoic</th>
<th>( A_{g, \text{max}} ) [mm²]</th>
<th>( A_{g, \text{min}} ) [mm²]</th>
<th>( A_{g, \text{max}}^{\text{tmax}} ) [mm²]</th>
<th>( A_{g, \text{max}} / A_{g, \text{min}} ) [-]</th>
<th>( A_{g, \text{max}}^{\text{tmin}} / A_{g, \text{min}}^{\text{tmax}} ) [-]</th>
<th>N*</th>
</tr>
</thead>
<tbody>
<tr>
<td>M-1st group</td>
<td>187 ± 9</td>
<td>185–198</td>
<td>194 ± 221</td>
<td>1.03 ± 0.07</td>
<td>1.01–1.08</td>
<td>14</td>
</tr>
<tr>
<td>F-1st group</td>
<td>186 ± 15</td>
<td>150–241</td>
<td>189 ± 162</td>
<td>1.01 ± 0.05</td>
<td>0.96–1.04</td>
<td>47</td>
</tr>
<tr>
<td>M-2nd group</td>
<td>199 ± 34</td>
<td>154–299</td>
<td>224 ± 348</td>
<td>1.15 ± 0.3</td>
<td>1.00–1.24</td>
<td>26</td>
</tr>
<tr>
<td>F-2nd group</td>
<td>198 ± 32</td>
<td>157–210</td>
<td>190 ± 246</td>
<td>1.31 ± 0.18</td>
<td>1.20–1.61</td>
<td>20</td>
</tr>
<tr>
<td>M-all</td>
<td>196 ± 32</td>
<td>154–299</td>
<td>217 ± 348</td>
<td>1.12 ± 0.24</td>
<td>1.00–1.24</td>
<td>59</td>
</tr>
<tr>
<td>F-all</td>
<td>177 ± 54</td>
<td>190–241</td>
<td>189 ± 274</td>
<td>1.08 ± 0.10</td>
<td>0.96–1.61</td>
<td>67</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Rapid Breathing–Tachypnoic</th>
<th>( A_{g, \text{max}} ) [mm²]</th>
<th>( A_{g, \text{min}} ) [mm²]</th>
<th>( A_{g, \text{max}}^{\text{tmax}} ) [mm²]</th>
<th>( A_{g, \text{max}} / A_{g, \text{min}} ) [-]</th>
<th>( A_{g, \text{max}}^{\text{tmin}} / A_{g, \text{min}}^{\text{tmax}} ) [-]</th>
<th>N*</th>
</tr>
</thead>
<tbody>
<tr>
<td>M-1st group</td>
<td>209 ± 10</td>
<td>200–217</td>
<td>214 ± 221</td>
<td>1.02 ± 0.04</td>
<td>0.97–1.05</td>
<td>41</td>
</tr>
<tr>
<td>F-1st group</td>
<td>179 ± 16</td>
<td>162–196</td>
<td>181 ± 230</td>
<td>1.00 ± 0.04</td>
<td>0.94–1.15</td>
<td>98</td>
</tr>
<tr>
<td>M-2nd group</td>
<td>214 ± 10</td>
<td>175–294</td>
<td>216 ± 219</td>
<td>1.11 ± 0.04</td>
<td>1.01–1.25</td>
<td>85</td>
</tr>
<tr>
<td>F-2nd group</td>
<td>175 ± 18</td>
<td>157–223</td>
<td>196 ± 252</td>
<td>1.17 ± 0.18</td>
<td>1.08–1.34</td>
<td>28</td>
</tr>
<tr>
<td>M-all</td>
<td>212 ± 10</td>
<td>175–294</td>
<td>228 ± 219</td>
<td>1.08 ± 0.09</td>
<td>0.97–1.25</td>
<td>126</td>
</tr>
<tr>
<td>F-all</td>
<td>178 ± 43</td>
<td>157–223</td>
<td>184 ± 252</td>
<td>1.04 ± 0.09</td>
<td>0.94–1.34</td>
<td>126</td>
</tr>
</tbody>
</table>

3.2.1. General features

Every mean breathing cycle of the database is characterized by a mean ratio \( A_{g, \text{max}} / A_{g, \text{min}} \) higher than 1. However, the time-variations of the detected glottal area demonstrate a substantial inter-subject variability. Within each gender and task category, the subjects were divided into two groups, represented in Table 3:

- The 1st group (labeled as “static”) gathers the subjects with a quasi-constant glottal area detected during the average breathing cycle. This group is characterized by a mean glottal area variation smaller than 10%. The limit was assessed as the accuracy of the detection procedure. This “static” glottal evolution during breathing has been reported in Hyatt et al. [39] and in 1 subject out of 3 in Jackson et al. [8].

- The 2nd group (labeled as “dynamic”) gathers the subjects with a time-varying glottal area during the average breathing cycle, so that the mean glottal area variation is higher than 10%. This group is characterized by a progressive glottal widening (resp. narrowing) during inspiration (resp. expiration), in agreement with previous reference studies [9,10,29]. In that case, the peak values reached during inspiration have been measured around 1.26 and 1.46 times higher than the minimum value achieved during expiration for males and females respectively. During slow breathing, the 2nd group is represented by 52.6% of subjects, against 44.4% during rapid breathing.

In our database, during slow breathing, the peak glottal area changes from \( A_{g, \text{max}} = 217 \pm 54 \text{ mm}^2 \) (mean ± SD) during inspiration to \( A_{g, \text{min}} = 178 \pm 35 \text{ mm}^2 \) during expiration, for males. Similarly, for females, the peak glottal area changes from \( 189 \pm 32 \text{ mm}^2 \) during inspiration to \( 168 \pm 31 \text{ mm}^2 \) during expiration. The mean value over the whole cycle, \( A_{g} \), was found at \( 196 \pm 42 \text{ mm}^2 \) for males, and \( 177 \pm 27 \text{ mm}^2 \) for females. The corresponding values for tachypnea breathing are stated in Table 3.

Another important feature can be highlighted from the present work. In average on our database, the measured glottal areas are much higher than those measured with equivalent method in other works [9,10]. Indeed, Baier et al. [9] measured a mean area \( A_{g} \) of 44 mm² and 48 mm² for 2 males. This area ranged from 43 mm² to 104 mm² for 4 females. Similarly, Brancatisano et al. [10] measured an area \( A_{g} \) of 98 mm² (\( A_{g, \text{max}} = 126 \pm 8 \text{ mm}^2 \), 12 subjects, 2 females).

However, our results are in agreement with previous studies based on the acoustic reflection [40–42] or MRI technique [43]. In particular, D’Urzo et al. [40] used both acoustic reflection and CT methods to measure glottal area of 11 subjects. The results derived from both methods were similar. The values of \( A_{g} \) were measured at functional residual capacity, and varied between 144 and 211 mm² on 25 males and between 137 and 207 mm² on 19 females during hyperpnea.

A glottal narrowing during expiration was observed in all studies. This is supported by the activity of the thyroarytenoid muscle during expiration [44], which is an adductor muscle of the vocal folds.

3.2.2. Influence of subject gender

The influence of the subject gender on glottal variations during slow breathing is clearly highlighted in Fig. 4. In the upper panel (Fig. 4a and b), it is shown that 9 male subjects have much more representatives in the 2nd group “dynamic” (7/9) relative to the 1st group “static” (2/9). On the contrary, the lower panel (Fig. 4c and d) shows that 10 female subjects are gathered together in the 1st group “static” (7/10), compared with the 2nd group “dynamic” (3/10). These
trends are similar in rapid breathing, as displayed in Fig. 5 (6/9 males against 2/9 females only in 2nd group “dynamic”). Despite their minimal proportion within the 2nd group, females demonstrate glottal motion amplitudes larger than males. This result yields to higher ratios $\frac{A_{g,\text{max}}}{A_{g,\text{min}}}$ and $\frac{A_{g, \text{max}}}{A_{g, \text{min}}}$ for females, whatever the task (see Table 3, darkened rows). For both genders however, the mean loops $\overline{A_g}(\overline{Q})$ plotted in Fig. 4b and 4d do not exhibit ellipsoidal shapes, which would have been observed in case of sinusoidal dynamics. Instead, the shape-differences observed on the loops $\overline{A_g}(\overline{Q})$ of males and females mainly derive from the different time-variations of the glottal area recorded during inspiration for both genders: a $\overline{A_g}$ plateau occurs during this phase for females, while a quasi-sinusoidal variation is measured for males (see Fig. 6). Note that during expiration, the glottal dynamics is similar for males and females: the glottis tends to stay opened in the narrowest posture (Fig. 6a and c).

### 3.2.3. Influence of breathing task

The comparison of the loops $\overline{A_g}(\overline{Q})$ given in Figs. 4 and 5 shows similar tendencies and orders of magnitude for eupnea and tachypnea.

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**Fig. 4.** Glottal area variations during slow breathing. Upper panels (a and b) are for male subjects, and lower panels (c and d) are for female subjects. Left panels (a and c) are dedicated to the 1st group (with “static” glottal movement) and right panels (b and d) are dedicated to the 2nd group (with “dynamic” glottal movement).

**Fig. 5.** Glottal area variations during rapid breathing. Upper panels (a and b) are for male subjects, and lower panels (c and d) are for female subjects. Left panels (a and c) are dedicated to the 1st group (with “static” glottal movement) and right panels (b and d) are dedicated to the 2nd group (with “dynamic” glottal movement). Panel a includes a detail of subject M08 for its high airflow rate variations.
From our results, the influence of the breathing task on glottal motion is far less emphasized than the one of the subject gender.

The glottal aperture is larger during the whole cycle for tachypnea than for eupnea. This is confirmed in Baier et al. [9], who measured larger glottal areas by both men and women during tachypnea than during quiet breathing.

3.3. Correlation between airflow-rate pattern and glottal motion

Fig. 6 shows the mean normalized glottal area dynamics ($A_g/A_{g,max}$) in function of normalized time ($\omega t$) for subjects classified within the 2nd group only. The 1st group is not presented here as the glottal dynamics is steady and the resulting normalized pattern would be a straight line close to 1. Upper (resp. lower) panels correspond to male subjects (resp. female). The mean normalized airflow-rate pattern ($Q/Q_{max}$) is also displayed, to analyze the correlation between airflow-rate and glottal area time-variations. For all 2nd group subjects and tasks, the glottal widening starts with the onset of inspiration, whereas the glottal narrowing precedes the expiration phase. For males, the peak glottal area $A_{g,max}$ is achieved after the peak inspiratory airflow $Q_{max}$, with a phase difference of 15.6°. For females, this delay is reduced and both peaks occur rather simultaneously. If the glottal motion seems related to the airflow dynamics during inspiratory phase, no correlation is brought out from the data registered during expiratory phase. Therefore, the same absolute airflow amplitude is associated with (at least) two different glottal openings.

4. Discussion

4.1. Toward the reasons for glottal motion and aerodynamics: an energy saving mechanism?

The respiratory-related glottal motion is generated by contractions of intralaryngeal muscles. The vocal-fold abductor (posterior criocartilaginous muscle) dilates the glottis during inspiration by abduction rotation of the arytenoid cartilages. The adductor muscles narrow the glottis during expiration by adduction-rotation of the arytenoid cartilages. There is general agreement that this controlled motion is carried out in order to (i) help the air intake during inspiration by glottis widening, which lowers the UA resistance to flow, and (ii) to slow down the passive expiratory airflow by glottis narrowing, which enables a control of the end-expiratory lung volume and prevents alveoli to collapse [10,45]. Note that several studies are dedicated to the restoration of an optimal laryngeal geometry after a surgical change of the vocal folds, which would minimize the airflow resistance in the UA [46,47].

Thus, the results obtained in our study on glottal motion and aerodynamics during breathing were further analyzed to interpret the data in terms of energy efficiency. To this end, the power dissipated by an airflow (volume airflow rate $Q(t)$) through a circular diaphragm ("glottis" of cross section area $A_g$) across a cylindrical tube ("trachea") of cross section area $A^t = A_{g,t}$ was calculated in two comparative cases:

(i) a realistic case mimicking the measured unsteady flow and glottal conditions, where $Q(t) = \bar{Q}(t)$ and $A_{g,t}(t) = A_{g,t}(t)$; and
(ii) an idealized case assuming unsteady sinusoidal flow conditions and steady glottis, where $Q(t) = Q_{max} \sin(\omega t)$ with $Q_{t,max} = \pi \bar{V}_t/\bar{T}$, and $A_{g,t} = \overline{A_g}$.

Considering the air as an incompressible Newtonian gas with typical flow Reynolds number $Re < 10^5$, the power $P_w$ [W] dissipated through a circular diaphragm is given by:

$$P_w = \Delta p \cdot Q^3 = \frac{1}{2} \rho \overline{U^2} \lambda \left| Q^0 \right|,$$

where $\Delta p$ [Pa] is the pressure drop through the singularity, $\rho = 1.2$ [kg.m$^{-3}$] is the air density, $\overline{U} = 2\bar{V}/(\bar{T}A_{g,t})$ [m.s$^{-1}$] is the mean flow velocity during inspiration phase and $\lambda [-]$, the singular head loss coefficient [48] calculated by $\lambda = (1 + 0.707\sqrt{1 - \frac{A_{g,t}}{A_{g,t}^0} - \frac{A_{g,t}^0}{A_{g,t}^0}})^2$. Fig. 7
The database was also exploited to evaluate the impact of the measured glottal motion and airflow rates on characteristic flow parameters within the glottis, listed below:

- The maximum and mean flow velocity during inspiration, noted $u_{\text{max}}$ and $u_{\text{mean}}$;
- The local Reynolds numbers $R_{\text{e, max}}$ and $R_{\text{e, mean}}$ based on $u_{\text{max}}$ and $u_{\text{mean}}$, respectively;
- The frequency parameter, given by $\alpha = R_h \sqrt{\omega / U}$, where $R_h = D_h / 4$, $D_h$ [m] being the hydraulic diameter and $\nu$ [m$^2$.s$^{-1}$] the air kinematic viscosity.

For each group, these parameters are function of the inspiratory volume $V_i$, the breathing period $T_t$ and the glottal cross-section area $A_g(t)$. Table 5 presents the values obtained in conditions of slow and rapid breathing for males and females (based on data from Tables 2 and 3). The mean Reynolds numbers range between 1900 and 3100 (peak values below 5000), corresponding to moderate mean velocity values expected within the glottis (peak values between 4 and 6 m.s$^{-1}$). Low values of the frequency parameter (in the range 1.8–2.7) are indicators of a flow comprising notable viscous effects, and whose time duration of the boundary layer development is comparable to the breathing cycle period. These parameters provide information, which could be used as input data to conduct further realistic numerical simulations of laryngeal airflow.

Finally, it is important to note, that the fraction of aerosols deposited on the laryngeal walls is highly dependent on the Reynolds number, defined as: $St = \rho_p d_p^2 u / (18 \mu D_h)$, where $\rho_p$ is the density of the particles, $d_p$ the particle diameter, $\mu$ the air viscosity, $u$ the air inlet velocity and $D_h$ the glottal hydraulic diameter. Thus, the deposition dynamics in the larynx is likely to be altered (i) by unsteady flow velocity, as demonstrated in [49], in case of static wall conditions, but also (ii) by unsteady glottal dimensions during breathing. From this point of view, the clinical database reported in this study seems to us of primary interest to better evaluate the aerosol transport and deposition mechanisms in future work.

## 5. Conclusion

A clinical study was conducted to examine the glottal motion during eight breathing tasks by means of laryngofibroscopy and synchronized oral airflow measurements on 20 healthy non-smoking volunteers from 20 to 30 year-old. In total, 144 breathing sequences of 30s were recorded. Several results can be highlighted:

Two groups of subjects were identified: (i) a “static” group, gathering subjects with a constant glottal area detected during the average breathing cycle; (ii) a “dynamic” group, gathering subjects with a time-varying glottal area during the average breathing cycle. This latter group gathers 53% (respectively, 44%) of subjects during slow (respectively, rapid) breathing.

### Table 5

| Flow parameter | Females | | Males | |
|----------------|---------|----------------|----------------|
| $u_{\text{max}}$ [m.s$^{-1}$] | 4.17 | 5.41 | 3.93 | 5.99 |
| $u_{\text{mean}}$ [m.s$^{-1}$] | 2.66 | 3.45 | 2.50 | 3.81 |
| $R_{\text{e, max}}$ [-] | 3069.2 | 4001.1 | 3087.9 | 4852.2 |
| $R_{\text{e, mean}}$ [-] | 1953.9 | 2547.2 | 1965.8 | 3089.0 |
| $\alpha$ [-] | 1.76 | 2.51 | 1.88 | 2.74 |
| $D_h$ [mm] | 10.7 | 10.8 | 11.44 | 11.8 |

shows a comparison of the power $P_w$ [mW] obtained from the data registered for female and male subjects, in the illustrative case of rapid breathing ($\text{Tachyp}_{30}$). Over the whole, it is interesting to note a substantial decrease of the power dissipated during the average breathing cycle in realistic case (i) as compared to idealized case (ii). The energy dissipated by head loss within the “glottal” diaphragm over a cycle, $E$ [J], is defined as $E = \int_0^{T_f} P_{\text{diss}}(t)\,dt$. Average values are presented in Table 4 in function of gender and breathing task. During slow breathing (resp. rapid breathing), the relative decrease between idealized and realistic cases equals 4% (resp. 6%) and 12% (resp. 13%) for males and females group respectively. Therefore, it is shown that the realistic breathing mode (mobile glottis, non-sinusoidal airflow rate) is an energy saving maneuver in comparison with an idealized mode (steady glottis, sinusoidal airflow rate).

### Table 4

| Energy loss $E$ [mJ] within a circular diaphragm in two breathing modes: (i) mobile glottis, non-sinusoidal airflow and (ii) steady glottis, sinusoidal airflow. |
|-----------------|-----------------|
| Slow breathing–E| Realistic case (i) | Idealized case (ii) |
| Males | 2.73 | 2.84 |
| Females | 2.73 | 3.10 |
| Rapid breathing–Tachyp$_{30}$ | Males | 5.28 | 5.62 |
| Females | 2.86 | 3.28 |
During "slow" breathing tasks, for males, the peak value of glottal area narrowed from 217 ± 54 mm² during inspiration, to 178 ± 35 mm² during expiration. For females, the peak glottal area changed from 189 ± 32 mm² during inspiration, to 168 ± 31 mm² during expiration. The mean value over the whole cycle was found at 196 ± 42 mm² for males, and 177 ± 27 mm² for females. These values are in agreement with previous studies based on acoustic reflection or MRI techniques. Yet, they are much higher than those previously determined by laryngofibroscopy, due to different spatial calibration.

A correlation motion related to the airflow dynamics was found during inspiratory phase, while no correlation was brought out from the data registered during expiratory phase.

Using a simple theoretical framework based on the power dissipated through a movable diaphragm, this study showed that the realistic breathing mode (mobile glottis, non-sinusoidal airflow rate) is an energy saving maneuver in comparison with an idealized mode (steady glottis, sinusoidal airflow rate).

Finally, this clinical database constitutes available input data which can be used to conduct further realistic numerical simulations of laryngeal airflow in realistic geometries of upper airways.

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Conflicts of interest

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