Low-Resistance Precision Vaporizers (*manuscript)*

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Summary

Draw-over vaporizers pose an alternative to plenum vaporizers in emergency situations and in remote areas despite the low precision of the former.

The instability of the existing draw-over vaporizers stems from the significant difference of flow regimes and geometry of bypass and vapor channels. This difference causes a dramatic dependence of the splitting ratio on the total gas flow. At low gas-flow rates through the vaporizer, the rate of chamber flow through control valves is comparable with the rate of secondary gas flows arising from non-uniform density, variations of pressure and instant velocity during artificial or spontaneous ventilation and because of other disturbing factors.

However, low hydraulic resistance is not necessarily related to low accuracy. An attempt to regularize the governing processes (gas flow, mass and heat transfer) within the laminar flow regime enabled us to design new low-resistance vaporizers (pocket vaporizer of 300 g mass, and a universal vaporizer), which are capable of dosing out anesthetics in the flow range of 0.2-15 $L \cdot min^{-1}$ just like plenum vaporizers.

The article considers characteristic features of portable anesthesia machines based on the above vaporizers, related to breathing circles, saving of anesthetics and gases, and auto-analgesia.

Keywords *Equipment:* low resistance (draw-over) vaporizers. *Anesthetic:* volatile. *Processes:* gas flow, mass- and heat transfer.

Introduction

"This has left something of a generation gap in the equipment available since draw-over vaporizer development all but stopped some 2 decades ago" **[1]**. Unfortunately, well-known draw-over vaporizers (*OMV, Ohmeda PAC, Goldman*) are associated with unpredictable output and inadaptability to low gas flows (below 4 $L \cdot min^{-1}$). At the same time, the most popular and rational flow range is from 0.2 to 4 $L \cdot min^{-1}$ (minimal environmental pollution and consumption of medical gases and anesthetics); low rate flows are also applicable in Pediatric and Veterinary anesthesia.

In spite of the gas flows non-stability, "draw-over anesthesia is the system of first choice for small hospitals" **[2, 3]**.

"An austere environment imposed by the tactical situation or geographical location may demand innovative approaches to what are normally routine clinical problems. For example, the scarcity of medical-grade compressed gas may require the anesthetist to use draw-over vaporizers … not in common practice in the U.S." **[4, 5]**.

Attributes of inhalation anesthesia system for remote area use or during major disasters [6]:

- Light weight, portable, comfortable, efficient, accurate, nonspillable;
- Providing for supplemental oxygen when available;
- Adaptable to various anesthetic agents;
- Ambient air can be used as carrier gas.

One might say that a perfect vaporizer would be as accurate as a plenum vaporizer and would have low-resistance as a draw-over vaporizer.

Analysis

Conditions of stability in *Plenum* and *Draw-over* vaporizers with dilution of the saturated vapor are similar:

- *Constant splitting ratio* of gas flow through the vaporizer,
- *Equilibrium saturation* of gas flow through the vapor chamber with anesthetic vapors,
- *Flow correction* by thermo compensator.

Constant splitting ratio would mean identical flow regime in the bypass and the vapor channels. In a low-resistance vaporizer, this takes place only in laminar flow **[7]**. First, the limited range of resistance (up to 10 mm H_2O) is used more rationally. This is so, because in laminar regime, pressure drop is proportional to the flow rate, while in turbulent regime it is proportional to the flow rate square. Second, when the vaporizer is in a breathing circle (VIC), the gas flow during inspiration of the patient instantly changes from zero to a maximum (about three times the minute ventilation when the inspiration and expiration phases are in ratio I:E = 1:2), and back. In such conditions, if the flow at low velocity is laminar, then it must be kept laminar at maximal velocity in both parallel lines, or else the stability of anesthetic dosage will deteriorate, which is easily demonstrable. Obviously, "the flow through either the bypass or the vaporizing channels should not change from laminar to turbulent (or vice-versa) at any point within the operating range of the vaporizer. If that did occur there would be a large and abrupt change in the calibration curve at that point " **[8]**.

Well-known draw-over vaporizers (*Goldman, OMV, Ohmeda PAC* and others) work just at non-stable *transient* flow mode and therefore cannot deliver stable anesthetic concentration at low gas flows.

Design Difficulties with Low-Resistance Vaporizers

Ensuring *constant splitting ratio* in low resistance vaporizers is much more difficult than in plenum vaporizers. First, the resistance of the control valves is about two orders of magnitude less (100 times), that is why at small flow rates (below 3 $L \cdot min^{-1}$), their control effect is comparable with the effect of disturbance factors which are of small importance for the plenum vaporizers. Second, the instant velocity, as noted above, varies in a wider range (from zero to maximal). At that, velocity variations further destabilize the splitting ratio, especially when there are considerable geometrical differences between the bypass and vapor channels (length, volume, configuration).

Example 1

Consider dependence of outlet isoflurane concentration on the total gas flow rate in the *Goldman* type vaporizer where:

- turbulent gas flow through the bypass channel has a rate from 3 to 12 $L \cdot min^{-1}$ and square-law dependence of pressure drop on the flow rate: $\Delta \mathbf{p_b = k_b F_b^2}$, where $\mathbf{k_b} = 0.011$ $Pa.min^2L^{-2}$ **[8]**;

- transient flow through the vapor channel (valve notch 1) has rate below 3 $L \cdot min^{-1}$ and pressure drop $\Delta \mathbf{p_c = k_c F_c^n}$, where, in first approximation $\mathbf{k_c} \approx 3.8$ $Pa.min^{1.5}$ $L^{-1.5}$ and the exponent is $\mathbf{n = 1.5}$;

- the vaporizing chamber delivers saturated isoflurane vapors with concentration $\mathbf{C_c = p_a p^{-1}} = 0.31$ or 31 vol. % (20°C, 760 mm Hg).

The delivered concentration from such vaporizer may be calculated using the formula:

$$\mathbf{C = [1 + (p\ p_a^{-1} - 1)\ F\ F_c^{-1}]^{-1}} \qquad \mathbf{(1)}$$

where the vaporizer flow rate is $\mathbf{F = F_b + F_c}$.

The calculation results are:

Bypass flow, $L \cdot min^{-1}$	3	6	12
Pressure drop of the bypass or chamber channels, Pa	0.1	0.4	1.6
Chamber flow F_c, $L \cdot min^{-1}$	0.09	0.22	0.56
Splitting ratio $F\ F_c^{-1}$	35.1	28.3	22.4
Outlet concentration C, vol. %	1.3	1.6	2.0

Thus, outlet concentration of the *Goldman* type vaporizer decreases approximately by 35% when gas flow rate decreases from 12 to 3 $L \cdot min^{-1}$ because of **different flow regime** through bypass and chamber channels (the exponent of pressure drop in the bypass channel is **n** = **2**, while the chamber exponent is **n** = **1.5**).

An additional fall of concentration can appear because of decreasing *mass transfer* in the vapor chamber. In that case, laminar stream of carrier gas passes strait from chamber inlet to outlet openings aside of evaporating surfaces, the very low rate of diffusion of agent vapor across the local streamlines above the liquid surface being a determining factor in the *Goldman* type vaporizer **[8]**.

Different density of carrier gas and anesthetic vapors is another disturbance factor for the draw-over vaporizers. At low flow rates of the carrier gas, say below 2 $L \cdot min^{-1}$, the pressure drop of the control valves is very low and comparable to the weight difference between the carrier gas and the anesthetic vapors.

Example 2

Evaluate outlet halothane concentration of *Oxford Miniature type Vaporizer (OMV)* at low flow rate. The vaporizer resistance is $\Delta\mathbf{p} = \mathbf{k_c}\cdot\mathbf{F_c^n}$ where $\mathbf{k} \approx 0.8$ and $\mathbf{n} \approx 1.5$ are experimental coefficients.

If the vaporizer pressure drop is $\Delta\mathbf{p}$ = 100 Pa at 25 L·min^{-1} (approximately equals to the bypass drop or the chamber drop), then $\Delta\mathbf{p}$ = 2.3 Pa at 2 L·min^{-1}.

Due to the pressure differential, part of carrier gas flow passes through the vapor chamber. However inlet orifices of the chamber outlet pipe are lower than the bypass axes by $\mathbf{h} \approx 60$ mm and there is a negative chamber pressure drop that is proportional to density difference of the carrier gas and the anesthetic (halothane) mixture:

$$\Delta\mathbf{p_d} \approx \mathbf{(\rho - \rho_a)\cdot g\cdot h} = (1.21 - 3.45)\ [\mathrm{kg.m^{-3}}]\ 9.8\ \ [\mathrm{m.s^{-2}}]\ 0.06\ [\mathrm{m}] = -1.3\ \mathrm{Pa}.$$

Thus, the actual chamber pressure differential is $\Delta\mathbf{p_c} = \Delta\mathbf{p} - \Delta\mathbf{p_d} \approx 2.3 - 1.3 = 1$ Pa, or about **40%** of the pressure drop. Accordingly, the chamber gas flow rate $\mathbf{F_c}$ and the outlet vaporizer concentration decrease, as follows from equation **(1)**. For example, at the scale mark **1%,** the outlet concentration will be only **0.4%,** and at **3%** - only **1.2%**.

This drop of concentration because of density non-uniformity is typical for *OMV, Ohmeda PAC* and *Goldman* vaporizers. The less is the gas flow through the vaporizer, the more of flow pressure differential is spent on the density pressure drop. In the Example 2, when carrier gas flow is 1.4 L·min^{-1}, heavy anesthetic vapors “shut up” the chamber which is disposed under the bypass channel.

Achievement of **equilibrium saturation** of gas flow through the vapor chamber should not interfere with the constant **splitting ratio**. Chambers of the well known *Vapor, TEC* or *PPV Sigma* vaporizers (long, deep, sharp bends) are not suitable for low resistance vaporizers because of the different geometry of the bypass (short and strait). In the case of

high-resistance (plenum) vaporizers, these differences do not stand out due to the insignificant relative resistance of the vapor chamber.

Recapitulation

The main reason for the output concentration non-stability of the well known draw-over vaporizers is the sharp dependence of *carrier gas splitting ratio* and *saturation coefficient* $\mathbf{s} = \mathbf{C_c \cdot p \cdot p_a^{-1}}$ on the total flow rate. One might interject the general comment that *splitting ratio* will be a more sensitive function of total flow rate at low values because of both different flow regimes of the parallel channels and some disturbing factors, namely:

- *Density fluctuations* (Example 2) because of non-uniformity of gas composition and temperature;

- *Variations of pressure* (see Appendix 3) and *instant velocity* during artificial and spontaneous ventilation;

- Diffusion of anesthetic vapors;

- Non-stability of vaporizer geometry (diameter and direction of inlet and outlet openings, cross-section of the vapor chamber at different anesthetic levels, etc).

"The overall process cannot be fully quantified …it is also clear that there are still fundamental mathematical difficulties with real fluids…" **[8]**.

However, the problem may be simplified if from the outset the object of exploration is free from the burden of imperfect devices and their functional dependencies which complicate the process. When splitting ratio and saturation coefficient are invariable at varying gas flow rates, it is not difficult to obtain from the Equation (1) simple relations between the output concentration, physical properties of the anesthetics and the control valves geometry, using known dependencies of the pressure drop in the parallel lines on the

mentioned factors **[7]**. Calculation error for the plenum vaporizer does not exceed 10% ("Anestezist-1") but grows significantly in the case of low-resistance vaporizers. These dependencies (not presented here for lack of space) may be used for control and adjustment of vaporizers, and for assessment of their operation under non-standard conditions (unusual temperature, pressure, carrier gas composition, etc.)

Low resistance is not necessarily related to low accuracy. Regularization of the governing processes (gas flow, mass and heat transfer) within the laminar flow regime enabled the design of new low-resistance vaporizers (a pocket vaporizer of 300 g mass, and a universal vaporizer) which are capable of dosing out anesthetics in the flow range of 0.2-15 $L \cdot min^{-1}$ just like plenum vaporizers.

Results

We have developed two different models of low-resistance vaporizers *FLOWVAP* **[9]** for expanding the area of their usage: pocket vaporizer *PV* **(Fig.1)** and universal vaporizer *UV*.

Fig.1

Main features of vaporizers:	*PV*	*UV*
Concentration range (isoflurane, halothane, enflurane, sevoflurane), vol. %	0 –3	0 – 6
Regulation	Stepped (0;0.5;1;2;3)	Continuous
Anesthetic volume, ml	30	100
Gas flow range, $L \cdot min^{-1}$	0.5 – 15	0.2 –15
Diameter of inlet/outlet taper connections, mm	15	22
Mass, kg	0.3	1.5

Hydraulic characteristics of the laboratory samples are shown in **Fig. 2**.

Fig.2

"Pressure drop – flow rate" dependence for the vapor channel was measured with the bypass channel closed, and vice-versa. The exponents of the dependence $\Delta \mathbf{p} = \mathbf{k} \cdot \mathbf{F}^{\mathbf{n}}$ for the parallel channels are approximately equal and increase proportionally from $\mathbf{n} \approx 1.1$ to $\mathbf{n} \approx 1.4$ with the flow rate (from 1 to 15 $L \cdot min^{-1}$) in which case the splitting ratio $\mathbf{F}\ \mathbf{F_c^{-1}}$ and the outlet concentration are approximately constant (**Table 1**).

Table 1

Technical data of *FLOWVAP* vaporizers and analogous devices are shown in **Table 2**.

Table 2

The concentration provided by *FLOWVAP* is virtually independent of the carrier gas flow rate when low and average concentrations are set, and falls when high concentrations are set together with high gas flow rate (**Fig. 3, 4)**. Output concentration as function of the ambient temperature is shown in **Fig. 5**.

Fig. 3, 4, 5

Essential Features of Anesthesia Machines with “FLOWVAP” Vaporizers

Breathing circles

Due to the low resistance (less than 10 mm H_2O) and virtual independence from the gas flow rate (0.2-15 $L \cdot min^{-1}$), *FLOWVAP* vaporizers may be both installed in conventional breathing circles (VIC) and used out of them (VOC), namely:

- in spontaneous and artificial respiration with air or oxygen O_2 and nitrous oxide N_2O with continuous and intermittent gas flows;

- with low pressure oxygen source (concentrator);

- with “to-and-fro” and circle absorption systems.

Installation

The simplest modification (see **Fig. 1**) is the pocket vaporizer *FLOWVAP* (mass 300 g, diameter 60 mm) that can be connected directly to a facemask or a tracheal tube by means of a non-rebreathing valve. This modification of the vaporizer should be available and used in emergency situations where, for example, amputation or disentanglement of victims from wreckage is required.

The minimal dead space, gas turbulence and flow resistance with accurate percentage of anesthetic vapors, even when low flow rates are used, would be relevant in pediatric anesthesia **[10]**.

Economy of anesthetics and medical gases

Approximately 60 ml of liquid anesthetic are retained after draining by wicks of the modern vaporizers *Vapor*, *Penlon SD*, *TEC 5*. In that case, the liquid anesthetic in wicks is blown off by flow rate 4 $L \cdot min^{-1}$ at the scale mark 4% for about 4 hours (see *Instruction for*

Use of Drager-Vapor 19.n) and the average concentration in case of isoflurane is ≈1,2 vol. % according to the mass balance equation.

On the other hand, short-time anesthesia (i.e. 30 min) needs about 7 ml isoflurane or 13 ml sevoflurane with carrier gas flow 4 $L \cdot min^{-1}$ when inspired anesthetic concentration is approximately equal to MAC (minimum alveolar concentration) in oxygen. Thus, 10-15 ml of anesthetic is enough for the short time anesthesia.

Auto Analgesia

Fig.6

The modification *ANESTAT-Auto* on **Fig. 6** can automatically regulate inspired anesthetic concentration depending on spontaneous ventilation. When the minute ventilation increases because of decreasing anesthesia depth or increasing surgical stimulation, inspired concentration increases too and vice-versa. When minute spontaneous ventilation decreases below the safety level set by the anesthetist, the patient breathes only fresh gas (oxygen or air + oxygen). This device actually "mimics" the function of an anesthetist, or rather assists him during maintenance of anesthesia at spontaneous breathing. In any moment, the anesthetist can manually regulate the concentration by the vaporizer scale. This method of anesthesia was approved **[11]**. Analogous research was carried out in Northwick Park Hospital, UK **[12]**.

Acknowledgments

I would like to thank Dr. Haim Halperin for managing the funding of the vaporizer development project, as well as Mr. Leib Berlin for managing the project and providing IT support.

Appendix 1

Notation

C	anesthetic concentration, vol. %
c	specific heat, $J.(kg.K)^{-1}$
D_a	diffusivity of anesthetic vapors, $m^2.s^{-1}$ ($cm^2.s^{-1}$)
d (r)	diameter (radius), mm
F	flow rate, $m^3.s^{-1}$ ($L.min^{-1}$)
f	ventilation frequency, min^{-1}
G	mass quantity, $kg.s^{-1}$
g	weight acceleration, $m.s^{-2}$
M	molecular mass, $g.mol^{-1}$
m	mass, kg
N	quantity
p	pressure, Pa (mm H_2O, mm Hg)
Q	heat quantity, W
S	surface, m^2
s	saturation coefficient
t	temperature, °C
V	porosity
u, v	linear velocity, $m.s^{-1}$
H, L, b, h, l	linear sizes, m (mm)
a, k	constants
μ	dynamic viscosity, Pa.s [$g.(cm.s)^{-1}$]
ρ	density, $kg.m^{-3}$

Appendix 2

Mass Transfer

The vapor chamber should provide *equilibrium saturation* of the passing gas flow by anesthetic vapors, under the additional conditions:

- *similar* hydraulic characteristics as the bypass (short, smooth, straight and minimal volume);

- *minimal residue of liquid anesthetic* after draining, ensuring minimal losses of costly anesthetics and pollution of the environment;

- adequate *heat exchange* with the ambient air.

Among well-known vaporizers, only *OMV* chamber with wire wicks matches these requirements except for the *similarity* of the bypass hydraulic characteristics. The last circumstance prevents accurate anesthetic delivery at low flow rates (see Example 2).

Fig. A1

The chamber of *FLOWVAP* vaporizers has longitudinal baffles **(Fig. A1)** with porous layer, forming parallel slit channels **[9, 13]**. The outlet vapor concentration in laminar gas flow is **[14]**

$$\ln [C_a (C_a - C)^{-1}] = 1.75\, \pi\, (D_a L F^{-1})^{2/3} \qquad \textbf{(A1)}$$

where $C_a = p_a p^{-1}$ is the concentration of anesthetic saturated vapors, D_a is the diffusivity of anesthetic vapors in carrier gas.

Example A1

Evaluate outlet anesthetic (halothane) concentration in the airflow passing through the vapor chamber. The chamber has $N = 10$ parallel slit channels (length $L = 15$ mm, width

b = 1.5 mm and height **H** = 30 mm). The airflow rate is **F** = 2 L·min^{-1} or 33.4 cm^3·s^{-1} at temperature **t** = 20°C.

Rewriting (**A1)** for one of the **N** rectangular channels we obtain:

$$\mathbf{\ln [C_a (C_a - C)^{-1}] = 1.75\ [(H + b)b^{-1}]\ (V\ D_a\ L\ N\ F^{-1})^{2/3}} \qquad \mathbf{(A2)}$$

where $\mathbf{C_a}$ = 0.323; $\mathbf{D_a}$ = 0.066 cm^2/s; and **V** = 0.4 is the porosity of the evaporating surfaces.

Thus, **C**= 0.313 or 31.3 vol. % which is 85 % from the saturated vapor concentration.

The experimental concentration at the exit of the vapor chamber was measured with Riken analyzer by diluting the outlet gas mixture (1:10):

Airflow rate $\mathbf{F_c}$, L·min^{-1}	0.5	1	2
Temperature of liquid halothane **t**, °C	≈22	≈23	≈17
Outlet halothane concentration **C**, vol. %	27.3	25.9	20.8
Saturation coefficient **s**, %	≈80	≈75	≈75 (≈**85***)

*** 85** is the calculated value in Example A1.

The above-cited high outlet concentrations are achieved at sufficient *heat feed* and *liquid anesthetic feed* to the porous evaporating surfaces.

Heat Transfer

There are three *heat sources* for evaporating the anesthetic: the first one is the *ambient air*, the second is the *carrier gas* and the third is the vaporizer *body specific heat*.

The necessary heat for evaporating of liquid anesthetic is determined from *conservation of energy equation* accounting for the carrier gas flow rate and the outlet anesthetic concentration.

Regarding the pocket vaporizer *PV*, due to its small mass (300 g) and relatively large consumption of heat, it cools down very rapidly to the dew point (approximately by a rate of 1°C per minute, at gas flow rate 10 $L \cdot min^{-1}$ and 3% concentration). Then, an additional source of heat appears from water vapor condensation in the ambient air convection flow. The *heat of condensation* compensates the shortage of heat for the anesthetic evaporation and slows the cooling of the vaporizer body until a stabilization temperature is achieved.

For example, if the ambient air temperature is 22°C and relative humidity is 85%, (water vapor partial pressure is then 2.25 kPa), the dew-point of the pocket vaporizer is 19.4°C and the stabilization temperature is $\mathbf{t}^* \approx 10$°C. The higher is the air humidity, the higher is the stabilization temperature of the vaporizer and vice-versa.

Temperature of evaporating surfaces

There are two heat flows towards the evaporating surfaces (see **Fig. A1**): *heat conduction* through the copper baffles and *heat convection* with the carrier gas flow passing through the chamber.

The chamber *heat convection* is an insignificant part (≈3%) of the necessary heat because of the great splitting ratio $\mathbf{F \cdot F_c^{-1}}$ (see **Table 1**). Therefore, temperature difference between the evaporation surfaces and the liquid anesthetic (vaporizer body temperature) is determined essentially by the *heat conduction*. Thus, in the above-cited example, the temperature of the evaporation surface is $\mathbf{t_s \approx t^* - \Delta t_c} \approx 7$°C at stabilization temperature $\mathbf{t}^* \approx 10$°C where the temperature drop $\Delta \mathbf{t_c} \approx 3$°C is determined from the standard heat-conduction equation for copper baffle plates with cross-section ≈10 mm^2. The larger the

vaporizer size, the less is the temperature difference between the evaporation surfaces and the ambient air. Thus, in the case of the universal vaporizer *UV*, the stabilization temperature is $\mathbf{t^*} \approx 15°C$ and the evaporating surface temperature is $\mathbf{t_s} \approx 12°C$.

Appendix 3

Influence of Pressure Fluctuations *(pumping effect of back pressure)*

Well-known studies have shown that the outlet vaporizer concentration during controlled or assisted ventilation is considerably higher than the outlet concentration when the vaporizer is used with free flow **[10]**. This difference is most pronounced when there is less anesthetic in the chamber, carrier gas flow is low, and pressure fluctuations are high and frequent.

Estimation of the pumping effect

Fig. A2

When artificial ventilation is carried out, pressure fluctuation $\mathbf{p_I}$ produced at the vaporizer outlet causes secondary gas portions $\mathbf{V_{gb}}$, $\mathbf{V_{gc}}$ with concentration **C** to move during *inspiration* from the outlet to a middle cross-section **O** of the chamber **(Fig. A2)**. During patient *expiration*, similar gas portions with concentration $\mathbf{p_a p^{-1}}$ move in opposite direction, from the middle cross-section to the vaporizer outlet. Due to these gas flows, pressure in the bypass and in the chamber follows the outlet pressure:

$$\mathbf{p_I\,(V_b + \Delta V_c) = p\,V_{gb}} \qquad \textbf{(A3)}$$

$$\mathbf{p_I\,(V_c - \Delta V_c) = p\,V_{gc}} \qquad \textbf{(A4)}$$

where $\mathbf{V_b + \Delta V_c}$, $\mathbf{V_c - \Delta V_c}$ are volumes of the bypass and chamber parts between the vaporizer outlet and the middle cross-section.

Thus, an additional amount of anesthetic vapors due to the *back pressure* is

$$\mathbf{V_a \approx (p_a p^{-1} - C)\,(V_{gb} + V_{gc})(1 - C)^{-1}} \qquad \textbf{(A5)}$$

Pressure drop of the parallel bypass or the chamber channel at *secondary laminar flow* is equal to

$$(k_b + k_{c1}) \cdot V_{gb} \approx k_{c2} \cdot V_{gc} \qquad \textbf{(A6)}$$

where k_b, k_{c1} and k_{c2} are resistance coefficients of the bypass valve, entry and exit chamber valves, respectively.

On the other hand, the above resistance coefficients can be found from similar equations for *fresh gas flow* through the bypass and chamber valves:

$$F\, F_c^{-1} = 1 + k_c k_b^{-1} [1 + \mu_m \mu^{-1} (1 - p_a\, p^{-1})] \qquad \textbf{(A7)}$$

where $k_{c1} = k_{c2} = k_c$; μ and μ_m are dynamic viscosities of gas and mixer of gas with saturated anesthetic vapors, respectively. Then the concentration increment is

$$\Delta C \approx V_a \cdot f \cdot F^{-1} \qquad \textbf{(A8)}$$

where **f** is the ventilation frequency.

Example A2

Evaluate the pumping effect for the pocket vaporizer *PV* at halothane concentration **C** = 1%; fresh gas flow rate **F** = 2 L·min^{-1}; frequency **f** = 15 min^{-1}; pressure fluctuation p_I = 200 mm H_2O (1,96 kPa); chamber and bypass volumes V_c = 66 cm^3, V_b = 2.4 cm^3 accordingly; **p** = 760 mm Hg (101,3 kPa), **t** = 20°C.

Using **(1),** after substitution of the specific values, we find:

$F \cdot F_c^{-1} = p_a\,(1 - C) C^{-1} (p - p_a)^{-1} = 46.5$;

From **(A7)**, $k_c k_b^{-1} = 15.5$; from **(A6),** $V_{gc} \cdot V_{gb}^{-1} = 1.06$.

Dividing **(A3)** by **(A4)** we find:

$\Delta V_c = (V_c - V_b\, V_{gc}\, V_{gb}^{-1})(1 + V_{gc} V_{gb}^{-1})^{-1} = 31$ cm^3 and

$\mathbf{V_{gb}} = 0.67\ cm^3$, $\mathbf{V_{gc}} = 0.7cm^3$.

Thus, the additional amount of anesthetic vapors is $\mathbf{V_a} \approx 0.43cm^3$ and the concentration increment is $\Delta\mathbf{C} = 0.0032$ or 0.32 vol. %.

Evaluated and experimental data for the back pressure are shown in **Table A1**. The concentration increment $\Delta\mathbf{C}$ is approximately *proportional* to the pressure fluctuation $\mathbf{p_I}$ and the chamber volume $\mathbf{V_c}$ but *inversely proportional* to the fresh gas flow rate **F**.

Table A1

Table 1

Dependence of the splitting ratio and the halothane concentration on the total airflow through the pocket vaporizer (calculated concentrations come from equation **(1)**, where

$\mathbf{F \cdot F_c^{-1} = (F \cdot F_c^{-1})_{Air} . \mu_m . \mu^{-1} (1-p_a p^{-1})^{-1}}$; $\mathbf{(F \cdot F_c^{-1})_{Air}}$ from **Fig. 2**; t=20 C, p=760 mm Hg)

Total air flow **F**, L.min^{-1}		1.5	4	9-10	12-13
Experimental splitting ratio	0.5	-	52	51	46
$\mathbf{(F \cdot F_c^{-1})_{Air}}$ at scale marks:	1	-	27	26	24
	2	11.4	13.1	12.7	-
Calculated concentration, vol. %	0.5	-	0.47 (100)	0.47 (99)	0.53 (88)
($\mathbf{F \cdot F_c^{-1}}$) at scale marks:	1	-	0.77 (51)	0.77 (51)	0.85 (46)
	2	2.1 (22)	1.8 (25)	1.9 (24)	-
Experimental concentration, vol. %	0.5	-	0.6	0.5	0.5
at scale marks:	1	-	1.15	1.1	1.1
	2	2.6	2.3	2.2	-

Table 2

Novel Low-Resistance Vaporizers and its Plenum and Draw-Over Analogs

Characteristics	PLENUM		DRAW-OVER		UNIVERSAL*	
	Vapor 2000	Penlon S Delta	OMV Penlon	PAC Ohmeda	FLOWVAP PV	FLOWVAP UV
Gas flow range, $L \cdot min^{-1}$	0.25-15	0.2-15	4-12	4-12	0.5-15	0.2-15
Temperature range, C	10-40	15-35	-	18-35	10-40	10-40
Anesthetic volume, mL	360	250	50	85	30	100
Wick volume, mL	60	60	10	35	5	5
Pressure drop at 15 $L \cdot min^{-1}$, mm H_2O	1100 (10 $L \cdot min^{-1}$)		10	15	10	10
Angle of tilt, degrees	30		30	90	180	180
Mass, kg	6.5-8.5	5.7	1.5	2.2	0.3	1.5

* FLOWVAP vaporizers can be used in either draw-over or plenum modes.

Table A1

Outlet concentration increment ΔC of the pocket vaporizer *PV* due to back pressure (ventilation frequency **f** = 15 min^{-1}; bypass volume $\mathbf{V_b}$ = 2.4 cm^3; **p** = 760 mm Hg, t = 20°C)

Pressure fluctuation p_I, mm H_2O	200	200	200	200	200	500	500
Flow **F**, L·min^{-1}	2	2	2	2	8	2	8
Chamber volume $\mathbf{V_c}$, cm^3	66	66	132	132	66	66	66
Set value of halothane concentration **C**, vol. %	1	3	1	3	1	1	1
Concentration increment Δ**C**, % vol	0.32 ≈0.3*	0.31	0.63 ≈0.85*	0.6 ≈0.3*	0.08	0.8	0.2 ≈0.1*

* experimental data

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