Diver respiratory responses to a tunable closed-circuit breathing apparatus

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Fothergill DM, Joye DD, Carlson NA. Diver respiratory responses to a tunable closed-circuit breathing apparatus. Undersea Hyperbaric Med 1997; 24(2):91–105.—Respiratory impedance of closed-circuit underwater breathing apparatus (UBA) is comprised of resistive, elastic, and inertial elements in series. Impedance is at a minimum when a UBA operates at its resonant frequency (f_n). This study investigated the respiratory responses of 12 male U.S. Navy divers to changes in the f_n of a simulated closed-circuit UBA. Respiratory effort, breathing comfort and ventilatory parameters were assessed during open- and closed-circuit breathing at rest and while exercising at 75 W on a bicycle ergometer in the dry at 1 atm abs. During closed-circuit breathing, the f_n of the system was adjusted to different frequencies between 0.2 Hz and 0.4 Hz (12 and 24 breaths/min) by varying UBA inertance. When the simulated UBA was switched from open to closed-circuit breathing the subjects changed their breathing frequency in a direction toward the f_n of the system and attempted to maintain minute ventilation constant by adjusting tidal volume. Results suggest that when divers breathe on a closed-circuit system with different f_n's they attempt to improve breathing comfort and reduce respiratory effort by adopting a breathing pattern that reduces their peak-to-peak mouth pressures.

diving, closed-circuit breathing apparatus, elastance, inertance, breathing impedance, respiratory effort

The total impedance to breathing of a closed-circuit underwater breathing apparatus (UBA) is comprised of resistive, elastic, and inertial elements in series. Resistance, or resistive impedance, results from the pressures required to move the breathing gasses through the CO_2 absorbent canister, valves, and breathing hoses at a particular flow rate. Elastance, or elastic impedance, arises from the elastic nature of breathing bags and the vertical displacement of the air–water interface as the volume of the breathing bag changes. Inertance, or inertive impedance, arises as a consequence of the mass of the breathing gas, water surrounding the bag, and the various UBA components being accelerated and decelerated with each breath.

Ideally, the design of a closed-circuit UBA should aim to minimize total impedance to reduce the respiratory pressures needed to produce a given oscillatory flow. Impedance is at a minimum when a UBA operates at its resonant frequency (f_n). At the f_n the impedance contributed by elastance and inertance cancel each other, resulting in resistance as the only source of impedance. The efficacy of using inertance to cancel elastance may be demonstrated empirically by adjusting UBA inertance so that the f_n of the closed-circuit system matches the diver's breathing frequency. When this occurs, respiratory pressures should be minimized.

Since a commercial tunable close-circuit UBA does not exist, we built an experimental apparatus to simulate a closed-circuit UBA in which it was possible to adjust the f_n of the system to within the normal physiologic range of breathing frequencies expected at rest and during light exercise. The general principles and means for doing this in an actual closed-circuit UBA are described in an issued patent (1). In an accompanying paper (2), Joye and Wechgelaer introduce the theory and provide a mathematical model describing the simulated, tunable closed-circuit UBA. Our paper examines the consequences of breathing on such a system on a diver's breathing pattern and respiratory sensations. In particular, we were interested in determining a diver's respiratory responses under conditions where the f_n of the closed-circuit system differed significantly from their normal breathing frequency (f_b). A further objective was to validate the mathematical model described in Joye and Wechgelaer (2) with our empirical observations.

MATERIALS AND METHODS

Subjects: Twelve male active duty U.S. Navy divers participated in the study. The age, height, and weight of these subjects were 32.1 ● 2.7 (SD) yr, 176.1 ± 8.3 cm, and 87.5 ± 11.9 kg, respectively. Three divers were ex-smokers and the remainder were non-smokers. Half the subjects had extensive experience diving on closed-circuit rebreathers (i.e., >40 dives); the other half had not dived previously on a closed-circuit rebreather.

The rebreathing circuit

A schematic diagram of the experimental setup is shown in Fig. 1. The closed-circuit rebreathing loop consisted of a mouthpiece with a two-way breathing valve (2700 Series, Hans Rudolph, Inc.; Kansas City, MO) connected via respiratory hoses to the top end of a vertically mounted rigid plastic cylinder (76.0 cm long, i.d. = 10.16 cm). The bottom end of the plastic cylinder was immersed in water and capped by a large-bore T-fitting with outlets of unequal length. The long and the short outlets were placed at the same depth in the water tank (center of T-fitting located approximately 49.0 cm below water level). The long outlet consisted of a 124.5-cm length of flexible but non-collapsible hose with an i.d. of 5.08 cm. The short outlet connected to the opposite end of the T-fitting was a 7.6 cm long ball valve bored out to an i.d. of 4.45 cm. The ball valve was housed in a casing having a length of 10.0 cm and internal diameter of 6.35 cm.

Sofnolime was placed in the closed-circuit loop to scrub the exhaled CO_2 from the system. A water trap was placed on the inspired side of the breathing circuit to prevent accidental aspiration of water while breathing on the system. It should be noted that under normal operation, excursions of the water column were maintained well below the lid of the water tank. The total air volume contained in the system was 12.8 liters.

System elastance: The rigid plastic cylinder simulated a counterlung in a submerged UBA and provided the elastic load in the system. As the diver breathes, gas is moved into or out of the cylinder and water is displaced. The difference in water levels between the interior and exterior of the cylinder results in a pressure difference. The magnitude of the pressure change for a given change in volume (elastance) is dependent on the cross-sectional area of the cylinder and can be calculated using the equations shown in Joye et al. (3). Using these equations, the elastance of the water column was calculated to be 12.34 $cmH_2O \cdot liter^{-1}$. Correcting the elastance of the water column for the extra volume in the hoses, CO_2 scrubber, and water trap, etc., gave a value of 10.8 $cmH_2O \cdot liter^{-1}$ for the total system elastance. This agreed closely with a system elastance of 10.4 ± 1 $cmH_2O \cdot liter^{-1}$, which was measured statically using the methods outlined in Joye et al. (3). When a diver is on the system, an additional allowance for lung volume must also be made. Using a value of 6 liters for lung volume brings the total system elastance to 10.1 $cmH_2O \cdot liter^{-1}$ when a diver is on the system.

System inertance: Inertance is the pressure relative to the acceleration or deceleration of the mass of water moving in the cylinder and flexible hose. The ball valve alters the inertance of the system by adjusting the amount (i.e., mass of water) directed to the outlets during breathing. The more water directed to the long outlet, the greater the inertance. By changing the inertance of the system it is possible to tune the breathing circuit to different natural frequencies. Details for calculating inertance and the mathematic relationship between inertance, elastance, and the resonant frequency for this system are provided in Joye and Wechgelaer (2).

With the ball valve in the closed position, inertance of the system is obtained by adding the water column inertance to the inertance of the flexible tube. This was calculated to be 6.36 $cmH_2O \cdot s^{-2} \cdot liter^{-1}$. With the ball valve fully open, we found experimentally that very little water passed through the flexible tube; essentially all of it went through the ball valve. Under these conditions the total inertance of the system was 1.56 $cmH_2O \cdot s^{-2} \cdot liter^{-1}$.

System resistance: Resistance (R) in the air circuit was determined by connecting a breathing machine to the mouthpiece and measuring the peak-to-peak pressures ($\mathbf{\Delta}Pm_{pp}$) generated during tidal volume (V_T) excursions of 0.75 liter at frequencies (f) between 8 and 32 breaths/min. During this procedure the relief valve at the top of the water column was open, allowing atmospheric air to enter and leave the breathing circuit, thus by-passing the water column. To conform to the model of reactance tuning outlined in Joye and Wechgelaer (2), resistance was calculated using the formula below:

$$R = \mathbf{\Delta}Pm_{pp}/V_T\, 2\pi f$$

A value of 1.3 ± 0.17(SD) $cmH_2O \cdot s^{-1} \cdot liter^{-1}$ was obtained for the resistance in the air circuit. The total resistance of the system including the tuning circuit was estimated by using the measured minimum $\mathbf{\Delta}Pm_{pp}$ obtained during a frequency sweep with the breathing machine. We calculated a value of 9.0 $cmH_2O \cdot s^{-1} \cdot liter^{-1}$ with the ball valve closed and a value of 4.7 $cmH_2O \cdot s^{-1} \cdot liter^{-1}$ with the ball valve fully open. This latter procedure assumes that at the point of minimum pressure, all the impedance in the breathing circuit is due to resistive loading. The R values obtained for the total system were those that were used in the mathematical model to calculate the natural frequencies.

Empirical determination of resonant frequency: Resonant frequencies were determined with the ball valve fully closed (90°) and fully open (0°) and at intermediate angles of 30°, 45°, and 60°. The valve settings were measured by a large protractor fixed to the top plate above the tank. Angles could be measure to ±3°. The frequency at the point of minimum pressure was used as a measure of the f_n of the system. This was obtained for each valve setting by performing a frequency sweep between 8 and 32 breaths/min using the breathing machine (V_T = 0.75 liter), while continuously monitoring peak-to-peak pressure at the mouthpiece.

With the valve closed, the f_n was measured to be 0.15 Hz (9 breaths/min) within an accuracy of ±0.02 Hz (±1 breath/min). With the valve open, the f_n was 0.28 Hz (17 breaths/min) within an accuracy of ±0.02 Hz (±1 breath/min). These values are lower than the undamped f_n's of 0.2 and 0.4 Hz (12 and 24 breaths/min) derived mathematically using the values of system inertance and elastance given above [Equation 9 in (2)]. The downward shift in frequency of minimum pressure is related to system resistance and occurs when f_n is determined at constant V_T. It does not occur when f_n is determined with constant minute ventilation (*see* Discussion).

Valve positions of 60° and 90° gave the same response, thus they both answer to a fully closed position. Valve openings of 30° and 45° gave minimum pressures at frequencies of 0.23 ± 0.03 Hz and 0.20 ± 0.02 Hz (14 ± 2 and 12 ± 1 breaths/min), respectively. Due to the complex flow patterns that occurred when the path of the water column was split between the short and the long outlets, a precise mathematical determination of the f_n could not be obtained for valve settings in between the fully closed and fully open positions.

Physiologic measurements

Mouth pressure relative to atmospheric pressure was monitored using a differential pressure transducer (±50 cmH_2O, Validyne Engineering Corp, Northridge, CA) mounted on a tap proximal to the orifice of the mouthpiece. Esophageal pressure relative to atmospheric pressure was measured with a balloon-tipped, air-filled catheter attached to a separate differential pressure transducer (±50 cmH_2O, Validyne Engineering). The catheter consisted of 1.4-mm i.d. polyethylene tubing, the distal end of which was pierced by small holes and covered with a 12 × 1.5-cm latex balloon filled with 2 ml of air. The balloon was inserted through the nostril and positioned in the middle third of the esophagus.

Gas samples were drawn from a separate tap on the mouthpiece and analyzed for end-tidal CO_2 and O_2 using an infrared CD-3A CO_2 analyzer (Applied Electrochemistry Ametek, Pittsburgh, PA) and a S-3A/I O_2 analyzer (Applied Electrochemistry Ametek), respectively. The speed of response to a step change in gas concentration was less than 100 ms to 90% of the final value for both gas analyzers. Gas samples were returned to the breathing circuit through a return sample line connecting the exhaust of the flow control to a manifold at the top of the vertical tube. A high pressure gas cylinder supplied 100% O_2 to the breathing circuit to maintain O_2 concentrations at normoxic levels while rebreathing. A manually adjusted flow control was used to ensure that the influx of O_2 matched the diver's O_2 consumption.

Inspired and expired volumes were measured using a S430A Ventilation Measuring System (K.L. Engineering, Northridge, CA). Heart rate (HR) was recorded using a 3-lead electrocardiogram system (Spacelabs 400, Redmond, WA).

A 386-IBM clone microcomputer with an A/D converter (model DAS-16F, Keithley/Metrabyte, Taunton, MA) sampled and stored mouth pressure, esophageal pressure, inspired volume, expired volume, and end-tidal CO_2 and O_2. Each channel was sampled at 60 Hz, stored on computer disk, and graphically displayed on-line using custom-designed software. The breath-by-breath data were analyzed off-line, using in-house software, to provide minute averages for peak inspired (Pm_{INSP}) and expired mouth pressures (Pm_{EXP}), peak-to-peak esophageal pressure (Pe), end-tidal CO_2 (PET_{CO_2}), expired minute ventilation ($\dot{V}E$), tidal volume (V_T), breathing frequency (f_b), inspiratory time (T_I), total breath time (T_T), and inspiratory duty cycle (T_I/T_T).

Experimental protocol

The experimental protocol was reviewed and approved by our institution's Committee for the Protection of Human Subjects. Volunteer subjects were informed of the nature of the experiment and provided informed consent; however, they remained blind to the switching of experimental conditions and were not instructed in any way on how best to breathe on the closed-circuit system. All data collection was performed with the subjects seated on an electromagnetically braked bicycle ergometer (Warren E. Collins, Inc., Braintree, MA) in the dry and at normal atmospheric pressure. During an experimental session the subject breathed on the simulated UBA first at rest and then while exercising at 75 W. A 15-min break was given between the rest and exercise session. During exercise, pedal rate was maintained at approximately 70 rpm; a time line of an experiment is shown in Table 1.

During both rest and exercise periods the closed-circuit system was tuned to resonant frequencies between 0.20 Hz (12 breaths/min, condition A) and 0.40 Hz (24 breaths/min, condition B) by adjusting the ball valve using the T-

Table 1: Time Line of Experiment

Minute	Operation
−2	Subject starts breathing on the simulated UBA in open-circuit mode at rest
−1	
0	Start computer data recording
1	
2	
3	Change simulated UBA to first experimental configuration (i.e., A, B, or C)
4	
5	
6	Get first RE and BC rating
7	Get second RE and BC rating
8	Change simulated UBA to second experimental configuration (i.e., A, B, or C)
9	
10	
11	Get first RE and BC rating
12	Get second RE and BC rating
13	Change simulated UBA to third experimental configuration (i.e., A, B, or C)
14	
15	
16	Get first RE and BC rating
17	Get second RE and BC rating
18	Change simulated UBA to experimental configuration D
19	
20	Get first RE and BC rating
21	Get second RE and BC rating
22	Change simulated UBA to experimental configuration E
23	Get first RE and BC rating
24	Get second RE and BC rating
25	Change simulated UBA to experimental configuration F
26	Get first RE and BC rating
27	Get second RE and BC rating
28	Stop data recording (subjects stop breathing on UBA circuit)
29	15-min break
44	Begin bicycle exercise at 75 W and repeat above procedure

bar shown in Fig 1. The handle of the T-bar was positioned so that it was aligned with a predetermined angle setting for each condition marked on the raised Plexiglas lid of the water tank. For open-circuit breathing (condition C), a relief valve at the top of the water column was turned to the open position so that subjects inhaled atmospheric air through the inspired respiratory hose and exhaled to the atmosphere through the CO_2 scrubber. When changing back from open-circuit to closed-circuit breathing, care was taken to ensure that the relief valve was closed during either mid-inspiration or mid-expiration. This ensured that the pressures required to move the water column were equally distributed between inspiration and expiration. A screen was placed in front of the subject so that they remained blind to the experimental conditions and apparatus.

At the start of each trial the subjects breathed on the open circuit for 5 min before being switched to the first experimental configuration. Conditions A, B, and C were presented in a random order, followed by conditions D, E, and F presented consecutively and in the same order for all subjects. Condition D_{Ex} during exercise, and F_R at rest were a repeat of conditions A and B, respectively, and were included as a check on the repeatability of responses within an experimental session. The remaining conditions were at intermediate valve positions between those of conditions A and B (rest; $D_R = 30°$, $E_R = 45°$; exercise, $E_{Ex} = 20°$, $F_{Ex} = 40°$) and were selected so that the range of resonant frequencies tested would coincide with the anticipated range of breathing frequencies expected during rest and exercise. Subjects repeated the experiment 1 wk later to assess the reproducibility of responses between trials.

Breathing comfort and perception of respiratory effort: As shown in Table 1, the subjects were asked to rate their breathing comfort (BC) and perceived respiratory effort (RE) twice during exposure to each condition. Breathing comfort was measured on a visual analog scale that consisted of an 18-in. ruler with anchors of *extremely comfortable* and *extremely uncomfortable* labeled at the far left- and right-hand ends, respectively. The anchor *extremely comfortable* was considered equivalent to normal resting breathing without breathing through a mouthpiece. Subjects were presented with the unmarked side of the ruler and asked to move a pointer, initially placed at the left side, to a position that best represented their current feelings of breathing comfort. Scores were recorded as the distance in inches from the left side of the scale. When rating BC, subjects were asked to consider the sensations listed in Table 2. This list was posted in front of the subject throughout the experimental trials.

Respiratory effort was described as a sensation analogous to that which accompanies the volitional act of lifting a weight (4). The subjects were instructed to be concerned only with the amount of effort it takes to breath in and out when rating RE and not with any other sensations associated with breathing. Respiratory effort sensations were rated using the Borg category-ratio scale shown in Table 3 (5). When using this scale, subjects were allowed to indicate ratings half-way between categories or numbers by pointing midway between two numbers. Subjects were also allowed to choose a number greater than 10 if their sensations of RE exceeded that which they experienced and rated as 10 in a previous experimental condition.

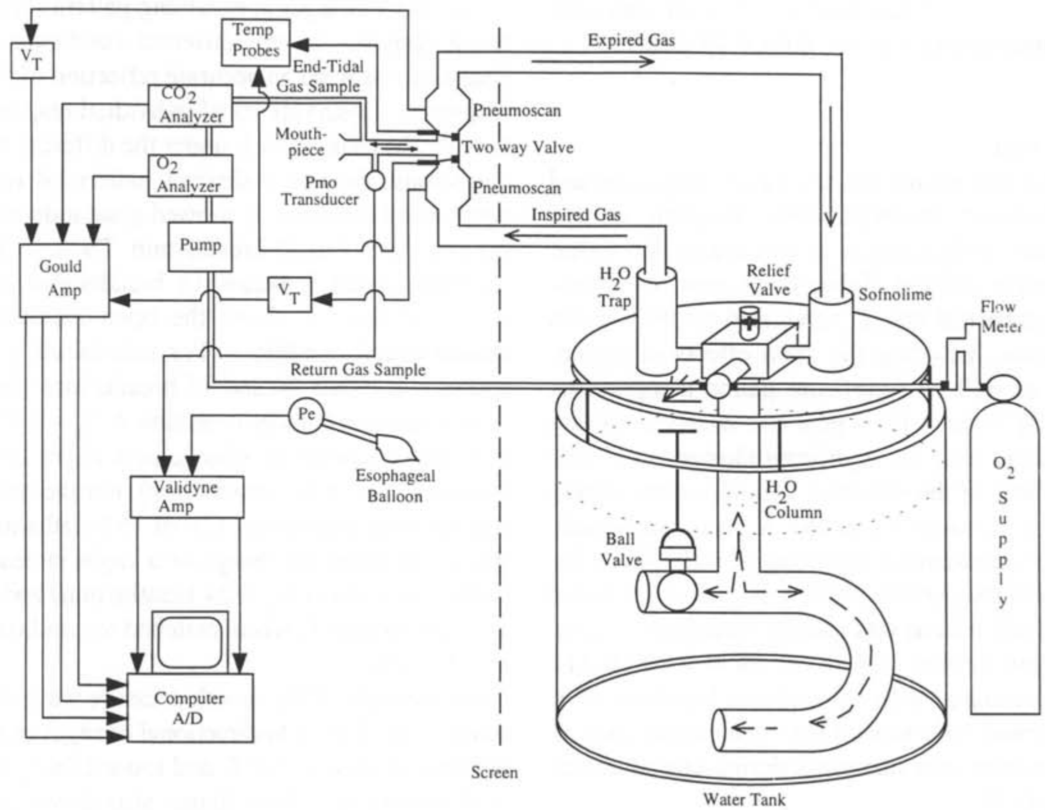

FIG. 1—Schematic diagram of the simulated closed-circuit UBA and experimental set up. The f_n of the closed-circuit system was manually adjusted by opening or closing the ball valve at the short outlet of the water column. System parameters with a diver breathing on the system with the ball valve fully open (0°) were: inertance (I) = 1.56, $cmH_2O \cdot s^{-2} \cdot liter^{-1}$; resistance (R) = 4.7 $cmH_2O \cdot s^{-1} \cdot liter^{-1}$; elastance (E) = 10.1 $cmH_2O \cdot liter^{-1}$; f_n = 0.40 Hz (24 breaths/min). System parameters with the ball valve fully closed (90°) were: I = 6.36 $cmH_2O \cdot s^{-2} \cdot liter^{-1}$; R = 9.0 $cmH_2O \cdot s^{-1} \cdot liter^{-1}$; E = 10.1 $cmH_2O \cdot liter^{-1}$; f_n = 0.20 Hz (breaths/min).

Table 2: Factors considered when rating breathing comfort on the visual analog scale

I feel that my breathing is too rapid
My breath does not go out all the way
My breath does not go in all the way
I feel that I am being smothered
I feel a hunger for more air
I cannot take a deep breath
My chest feels tight or constricted
I feel that I am suffocating
I am gasping for breath
My breathing is shallow
I feel that I am breathing too slow
I cannot get enough air
My breathing requires more concentration
My cheeks puff out
I feel pressure on my ears (ear squeeze)

Table 3: The Borg Category–Ratio Scale Used for Rating Respiratory Effort

0	None at all
0.5	Extremely slight (just noticeable)
1	Very slight
2	Slight
3	Moderate
4	Somewhat severe
5	Severe
6	
7	Very severe
8	
9	
10	Extremely severe (almost max)
•	Maximum

Data analysis

The first and second ratings for BC and RE for each breathing configuration were averaged before being subjected to the main statistical analysis. The physiologic responses corresponding with these ratings were taken from the average response over the minute immediately before the ratings were given. Repeated measures analysis of variance was used to determine the influence of test session and system configuration on the respiratory and psychophysical responses for the rest and exercise conditions. When appropriate, Tukey's HSD post hoc analysis was used to identify significant differences in responses to the various system configurations using the combined data for the initial and repeat test sessions. For all statistical tests a priori significance was set at the 0.05 level.

RESULTS

Responses at rest

Group means (±SEM) for the respiratory responses and breathing sensations observed while breathing on the different system configurations at rest during the repeat trials are shown in Table 4. Table 4 also presents a summary of the statistical results conducted on the various resting responses, including the main effects of system configuration as well as significant paired comparisons found following Tukey's HSD post hoc analysis. During the first trial there were no significant changes in f_b with alterations in the f_n of the system at rest. However, during the second trial, f_b under condition A was significantly greater than for open-circuit breathing ($P < 0.05$) and for closed-circuit breathing under conditions E_R ($P < 0.05$) and F_R ($P < 0.05$). Tidal volume was greater while breathing on the closed-circuit system with the f_n set at 0.20 Hz (12 breaths/min) compared with open-circuit breathing ($P < 0.05$) but decreased back toward the open circuit value as the f_n of the system was increased during closed-circuit breathing (Table 4).

Although the main effect of system configuration was significant for PET_{CO_2} ($P = 0.042$) and $\dot{V}_E$ ($P < 0.01$), Tukey's post hoc analysis did not find any significant pairwise comparisons for PET_{CO_2} and revealed that the only significant change for $\dot{V}_E$ was between conditions B and D_R ($P < 0.05$). Inspiratory duty cycle increased from 0.45 to approximately 0.50 when breathing was switched from open- to closed-circuit (C vs. all other comparisons, $P < 0.01$), but showed little change with exposure to the different closed-circuit configurations. Since inspiratory duty cycle remained constant at 0.50 the average mean pressure at the mouth was calculated from the sum of Pm_{INSP} and Pm_{EXP} to determine if inspiratory and expiratory pressures were divided equally during the different closed-circuit conditions. The largest deviation from zero in the average mean mouth pressure occurred for condition A (-3.3 cmH_2O). For the remaining conditions, the average mean mouth pressure varied between +2.1 and -1.6 cmH_2O. Pe was unaffected by the changes in the f_n of the system during closed-circuit breathing ($P > 0.05$) but was approximately twice as great as that observed during open-circuit breathing (C vs. all other comparisons, $P < 0.001$). A similar pattern of results was shown for Pm_{INSP} as for Pe. In contrast, Pm_{EXP} was twice as great with the ball valve in the fully closed position (conditions B and F_R) than when fully open (condition A).

The above changes in breathing pattern reflect the group mean response to the different conditions and do not necessarily provide an accurate reflection of the individual responses. Observations of individual responses, particularly for the changes in f_b under the different closed-circuit conditions, showed a distinct pattern of results. In the open-circuit condition, f_b showed great individual variation ranging from 7 to 21 breaths/min. Those individuals with an open-circuit f_b below 11 breaths/min (six subjects) increased their f_b above the open-circuit f_b during all closed-circuit conditions. For individuals with an open-circuit f_b between 14 and 16 breaths/min (three subjects) their f_b increased under condition A (f_n = 24 breaths/min) and showed either no change or a slight decrease under condition B (f_n = 12 breaths/min). For the remaining three subjects with open-circuit f_b's of 16 breaths/min or greater there was either no change or a slight increase in their f_b under condition A (f_n = 24 breaths/min) and a consistent decrease in their f_b when switched to condition B (f_n = 12 breaths/min).

An example of the mouth pressure wave forms for one subject who showed bi-directional changes in f_b away from his natural open-circuit f_b and toward the f_n of the system is shown in Fig. 2. This figure also shows the pressure at the mouth for three complete breathing cycles for one subject during open-circuit breathing and during closed-circuit breathing under conditions A and B. Note that in Fig. 2 an increase in f_b is denoted by a reduced time for the three breaths and a decrease in f_b by an increase in the time to complete the three breaths. The values for f_b shown for the different conditions in Fig. 2 were obtained from the average f_b taken over the final 2 min for each condition.

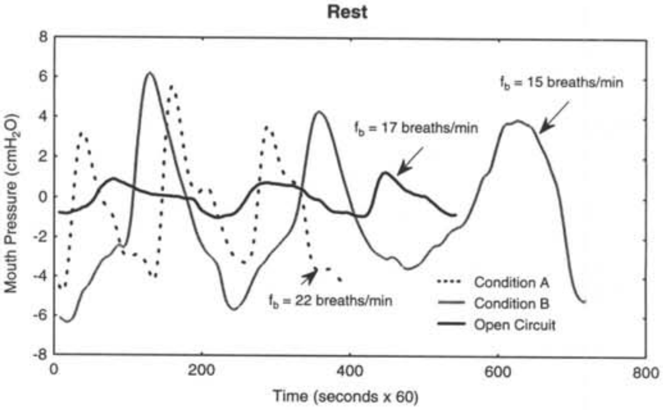

FIG. 2—Pressure at the mouth for three complete breathing cycles recorded from one of the subjects during open-circuit breathing and during closed-circuit breathing under condition A (f_n = 24 breaths/min) and B (f_n = 12 breaths/min) while at rest. To obtain time in seconds, divide the time axis value by 60. The average f_b over a 2-min period during the repeat trials is shown for each condition. Note the change in f_b away from the open-circuit f_b of 17 breaths/min and toward the f_n of the system during closed-circuit breathing.

Responses during exercise

Table 5 shows group means (±SEM) for the respiratory and psychophysical responses while breathing on the simulated UBA under the different conditions during light exercise. A statistical summary for the main effects of system configuration and Tukey's post hoc analysis is also included in Table 5 in the same form as in Table 4. The data shown are for the repeat trials. During exercise there was a significant and progressive decline in f_b and a concomitant increase in Pe, Pm_{EXP} and Pm_{INSP} as the f_n of the system was decreased from 0.40 to 0.20 Hz (24–12 breaths/min). The mean f_b while breathing on the open-circuit was not, however, significantly different from the mean f_b under condition D_{Ex}. Tidal volume increased progressively as the f_n was decreased but did not differ significantly from the open-circuit condition until the system was configured to the lower f_n's (i.e., B and F_{Ex}).

During exercise, $\dot{V}_E$ was lower under all closed-circuit configurations compared with open-circuit breathing (C vs. all other comparisons, $P < 0.01$). Further small but significant decreases in $\dot{V}_E$ (between 2 and 5%) were observed when the f_n of the system was decreased from 0.40 to 0.20 Hz (24–12 breaths/min) ($P < 0.05$). The only significant changes in T_I/T_T were between closed-circuit breathing with f_n= 0.40 Hz (24 breaths/min) (i.e., A and D_{Ex}) and condition F_{Ex}. The average mean mouth pressure tended to be slightly negatively biased for all conditions deviating from zero cmH_2O by between −0.9 and −2.8 cmH_2O. The only significant increase in PET_{CO_2} occurred when the subjects were switched from open-circuit breathing to the closed system with the lowest f_n (i.e., condition C vs. B, $P < 0.05$). Heart rates tended to be higher during the last three breathing configurations tested (i.e., D_{Ex}, E_{Ex}, and F_{Ex}) compared to conditions A, B, and C (Table 5), however, there were no significant differences in exercising HR between conditions A, B, and C.

In the open-circuit mode, individual values for f_b ranged from 19.4 to 30 breaths/min with the majority of subjects (9 out of 12) exhibiting exercising f_b's that were equal to or greater than the highest resonant frequency condition (i.e., 24 breaths/min for condition A). The subject with the lowest open-circuit f_b (19.4 breaths/min) increased his f_b to 21.1 breaths/min while breathing on the closed-circuit system under condition A. In contrast, the remaining subjects with higher open-circuit f_b's showed either no change or a decrease in their f_b when exposed to condition A. All subjects decreased their f_b below their open-circuit f_b during exposure to the remaining closed-circuit conditions B, D_{Ex}, E_{Ex}, and F_{Ex}. Figure 3 shows an example of the mouth pressure wave forms and changes in f_b observed in one subject when breathing was switched between the open-circuit configuration and closed-circuit condition A (f_n = 24 breaths/min) and B (f_n = 12 breaths/min) during exercise. The data are in the same form as in Fig. 2 and are typical of the pressure wave forms and changes in f_b seen in the majority of subjects when breathing on the simulated UBA under the different system configurations during exercise.

Respiratory effort and breathing comfort

Analysis of variance indicated a significant main effect of system configuration on RE and BC both at rest (Table 4) and during exercise (Table 5). The lowest ratings for RE and BC were obtained when breathing on the open circuit. When breathing was switched to the closed circuit, BC and RE increased significantly both at rest and during exercise (Tables 4 and 5). Changing the f_n of the system at rest did not significantly affect BC or RE. During exercise, ratings of RE increased by 40% ($P < 0.05$) and breathing became more uncomfortable ($P < 0.01$) when the f_n of the system was decreased from 24 to 12 breaths/min.

After the experiments, the divers' subjective reports indicated that a considerable increase in concentration was required to breathe on the system. Many divers noted that they consciously changed their breathing pattern in order to "breathe with the 'machine' rather than fight it." One diver commented that he didn't like the fact that he could not "skip breathe" (i.e., perform post-inspiratory or post-expiratory pauses) while breathing on the closed circuit.

Reproducibility of respiratory and psychophysical responses

We found no significant differences ($P > 0.05$) between test session 1 and 2 for $\dot{V}_E$, V_T, f_b, T_I/T_T, PI_{O_2}, PET_{CO_2}, Pm_{INSP}, Pm_{EXP}, and Pe during the rest and exercise conditions. However, a significant interaction between test session and system configuration was observed for f_b at rest ($P < 0.05$). Post hoc analysis revealed greater breathing frequencies during the repeat test session for system configurations A (12.4 vs. 14.8 breaths/min; $P < 0.01$) and B (10.9 vs. 13.1 breaths/min; $P < 0.01$). For system configurations C through F, breathing frequencies during test session 1 and 2 were the same ($P > 0.05$). Heart rates recorded at rest while breathing on the different system configurations were the same during the initial and repeat trials ($P > 0.05$); however, exercising HRs were on average 8 beats/min lower during the repeat trials ($P < 0.05$).

During exercise there was no difference in BC ($P > 0.05$) and RE ($P > 0.05$) between sessions 1 and 2; however, at rest BC was rated 35% lower (i.e., more comfortable) ($P <$

Table 4: Psychophysical and Respiratory Responses at Rest While Breathing on the Closed-Circuit System Tuned to Different Resonant Frequencies (f_n)

Condition	A	B	C	D_R	E_R	F_R	
Ball valve position	0°	90°	NA	30°	45°	90°	
f_n, Hz	0.40	0.20	NA			0.20	Significant Post Hoc Paired Comparisons, $P < 0.05$
f_n, breaths/min	24.0	12.0	NA			12.0	
f_b, breaths/min[a]	14.8±1.2	13.1±0.6	12.7±1.2	13.4±1.0	12.9±0.6	12.8±0.8	AC, AE_R, AF_R
V_T, liter, BTPS[‡]	1.24±0.08	1.39±0.12	1.13±0.07	1.06±0.09	1.16±0.12	1.34±0.14	AB, BC, BE_R, CF_R, D_RF_R
$\dot{V}_E$, liters/min, BTPS[†]	18.2±2.2	18.4±2.2	15.4±1.8	14.7±2.0	15.1±2.2	18.0±2.7	BD
Pe, cmH_2O[‡]	17.2±2.5	19.2±2.7	10.8±1.2	17.6±2.3	19.1±2.5	20.3±3.3	AC, BC, CD_R, CE_R, CF_R
Pm_{INSP}, cmH_2O[‡]	−8.1±0.6	−7.8±1.1	−1.2±0.1	−7.2±0.8	−7.2±0.9	−8.2±1.2	AC, BC, CD_R, CE_R, CF_R
Pm_{EXP}, cmH_2O[‡]	4.8±0.7	9.9±1.8	1.1±0.1	5.6±0.7	6.6±1.3	8.9±1.3	AB, AC, AF_R, BC, CD_R, CE_R, CF_R
T_I/T_T[‡]	0.51±0.020	0.50±0.008	0.45±0.019	0.51±0.018	0.49±0.015	0.49±0.014	AC, BC, CD_R, CE_R, CF_R
PET_{CO_2}, mmHg[*]	30.6±2.2	29.5±2.2	32.9±2.0	33.4±2.1	33.9±2.3	32.2±2.3	
HR, beats/min	73±3.7	75±4.6	75±4.7	73±4.0	74±3.3	73±3.2	
RE, Borg scale[‡]	1.56±0.37	1.63±0.32	0.46±0.78	1.56±0.32	1.53±0.34	1.55±0.35	AC, BC, CD_R, CE_R, CF_R
BC, VAS[‡]	3.46±0.85	3.44±0.83	0.87±0.39	3.37±0.71	3.41±0.78	3.56±0.80	AC, BC, CD_R, CE_R, CF_R

Key: f_b = breathing frequency; V_T = tidal volume; $\dot{V}_E$ = minute ventilation; Pe= Pk-to-pk esophageal pressure; Pm_{INSP} = peak inspired mouth pressure; Pm_{EXP} = peak expired mouth pressure; T_I/T_T = inspiratory duty cycle; PET_{CO_2} = end-tidal PCO_2; HR = heart rate; RE = respiratory effort rating; BC = breathing comfort rating; NA = not applicable. Ball valve position 0° = ball valve fully open, 90° = ball valve closed, condition C is open-circuit breathing. The data are group means ±SEM (n = 12, except for Pe where n = 11) for the repeat trial. Superscript symbols show significant main effect for system configuration following analysis of variance (* = $P < 0.05$; † = $P < 0.01$; ‡ = $P < 0.001$).

[a] Although the main effect of system configuration was not significant for f_b, there was a significant interaction between test sessions and system configuration ($F_{5,55} = 2.53$; $P < 0.05$). Analysis of this interaction showed no effect of system configuration on f_b during the first trial. The significant ad hoc paired comparisons shown for f_b above are for the repeat trial.

Table 5: Psychophysical and Respiratory Responses During Bicycle Exercise (Workload = 75 W) While Breathing on the Closed-Circuit System Tuned to Different Resonant Frequencies (f_n)

Condition	A	B	C	D_{Ex}	E_{Ex}	F_{Ex}	Significant Post Hoc Paired Comparisons ($P < 0.05$)
Ball valve position	0°	90°	NA	0°	20°	40°	
f_n, Hz	0.40	0.20	NA	0.40			
f_n, breaths/min	24.0	12.0	NA	24.0			
f_b, brths/min‡	21.9±0.7	15.5±0.3	24.2±0.8	22.5±0.5	20.0±0.5	18.0±0.4	AB, AC, AF_{Ex}, BC, BD_{Ex}, BE_{Ex}, CE_{Ex}, CF_{Ex}, DE_{Ex}, DF_{Ex}
V_T, liter, BTPS‡	2.17±0.07	2.90±0.11	2.20±0.09	2.15±0.07	2.28 0.08	2.48±0.09	AB, AF_{Ex}, BC, BD_{Ex}, BE_{Ex}, BF_{Ex}, CF_{Ex}, DF_{Ex}, EF_{Ex}
V_E, liters/min, BTPS‡	46.8±1.3	45.8±1.5	51.2±1.7	47.9±1.1	45.7±1.2	44.4±1.2	AB, AC, AF_{Ex}, BC, CD_{Ex}, CE_{Ex}, CF_{Ex}
Pe, $cmH_2O^{\ddagger}$	33.6±1.8	47.5±2.9	20.4±1.2	33.0±1.5	38.9±1.7	46.3±2.6	all comparisons significant except AD_{Ex} and BF_{Ex}
Pm_{INSP}, $cmH_2O^{\ddagger}$	−11.9±0.7	−23.5±1.3	−3.1±0.3	−12.1±1.0	−15.7±0.7	−20.8±1.1	all comparisons significant except AD_{Ex}
Pm_{EXP}, ($cmH_2O^{\ddagger}$	10.8±1.2	22.6±1.9	3.2±0.3	11.1±0.8	12.9±1.0	18.3±1.4	all comparisons significant except AD_{Ex}, AE_{Ex} and $D_{Ex}E_{Ex}$
$T_I/T_T^{\dagger}$	0.49±0.019	0.47±0.014	0.47±0.012	0.48±0.017	0.47±0.014	0.47±0.008	AF_{Ex}, $D_{Ex}F_{Ex}$
$PET_{CO}2$, mmHg*	42.6±0.9	42.9±1.0	41.1±0.9	41.5±0.9	42.6±0.9	42.9±1.0	BC
HR, beats/min*	124±5.5	123±5.6	124±5.4	129±5.4	130±5.8	131±5.0	AE_{Ex}, AF_{Ex}, BD_{Ex}, BE_{Ex}, BF_{Ex}, CD_{Ex}, CE_{Ex}, CF_{Ex}
RE, Borg scale‡	2.41±0.37	3.39±0.34	1.19±0.40	2.77±0.26	3.06±0.39	3.31±1.73	AB, AC, AF_{Ex}, BC, CD_{Ex}, CE_{Ex}, CF_{Ex}, DF_{Ex}
BC, VAS‡	5.12±0.83	7.00±0.87	2.23±0.71	6.25±0.75	6.63±0.98	7.57±1.16	AB, AC, AF_{Ex}, BC, BD_{Ex}, CD_{Ex}, CE_{Ex}, CF_{Ex}, DF_{Ex}

Key: *See* Table 4 for list of symbol definitions. Ball valve position 0° = ball valve fully open, 90° = ball valve closed, condition C is open-circuit breathing. The data are group means ± SEM (n = 12, except for Pe where n = 11) for the repeat exercise trial. Superscript symbols show significant main effect for system configuration following analysis of variance ($^{*} = P < 0.05$; $^{\dagger} = < 0.01$; $^{\ddagger} = P < 0.001$).

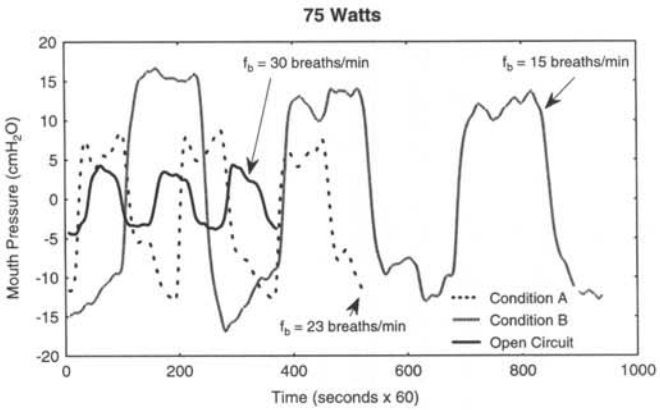

FIG. 3—Pressure at the mouth from three complete breathing cycles recorded from one of the subjects during open-circuit breathing and during closed-circuit breathing under condition A (f_n = 24 breaths/min) and B (f_a = 12 breaths/min) during exercise. Data are presented in the same form as for Fig. 2, with the average f_b over a 2-min period during the repeat trials shown for each condition. Note the decrease in f_b from the open-circuit f_b of 30 breaths/min toward the f_n of the system during closed-circuit breathing.

0.01) and RE 31% lower ($P < 0.05$) during the repeat session. Significantly lower RE and BC ratings during session 2 were observed only for the closed-circuit conditions and not for open-circuit breathing. Within an experimental session there were no significant differences in any of the respiratory responses or psychophysical ratings when the repeat conditions B and F_R at rest and A and D_{Ex} for exercise were compared.

DISCUSSION

Breathing pattern responses to reactance tuning

At rest, the effect of reactance tuning on breathing pattern was dependent on the individual's natural (open circuit) f_b relative to the f_n of the system. Those individuals who had a natural f_b close to the f_n of the system showed very little change in their breathing pattern. In contrast, individuals with a natural resting f_b either higher or lower than the closed-circuit f_n tended to change their f_b in a direction toward the f_n of the system.

During exercise, most subjects had open-circuit f_b's that were either the same or greater than the highest f_n condition for the closed-circuit system. Consequently, all subjects tended to reduce their f_b as the f_n of the closed circuit was decreased from 0.40 Hz (24 breaths/min) to 0.20 Hz (12 breaths/min). Progressively lowering the f_n of the system below the subject's open-circuit f_b resulted in breathing becoming more uncomfortable. This discomfort was associated with the increase in respiratory effort required to overcome the increase in breathing impedance in theclosed-circuit when more water was directed through the long outlet of the tuning system.

Figure 4 shows the consequences of changing the closed-circuit system configuration between conditions A and B on the predicted peak-to-peak mouth pressures generated when $\dot{V}_E$ is maintained constant at 46 liter/min. Minute ventilation isopleths at 46 liter/min are displayed, as this was the mean $\dot{V}_E$ observed during exercise under the closed-circuit conditions (Table 5). The curves in Fig. 4 were derived using the mathematical model (Equation 7) described in Joye and Wechgelaer (2). A comparison of the tuned curves A and B in Fig. 4 illustrates that for breathing frequencies greater than 10 breaths/min the respiratory cost of maintaining ventilation constant in terms of ΔPm_{pp} is greater in the latter compared to the former condition. The greater respiratory pressures required for condition B are due largely to the increases in inertance and flow resistance in the tuning circuit that occurs when the apparatus is tuned to the lower resonant frequency.

Results during exercise indicated that subjects preferred to lower their f_b toward the f_n of the system and attempt to maintain $\dot{V}_E$ by increasing V_T rather than generate the high respiratory pressures required to defend their f_b at the open-circuit level. Although larger ΔPm_{pp} need to be generated to breathe at a higher V_T during closed-circuit breathing, the decrease in ΔPm_{pp} obtained by decreasing f_b is much greater than the increase in ΔPm_{pp} resulting from the increase in V_T. From Fig. 4 it can be seen that if subjects had chosen to maintain their f_b at the open-circuit level (i.e., 24.2 breaths/min) while breathing on the simulated UBA under condition B, they would have to generate peak-to-peak mouth pressures well in excess of 60 cmH_2O to meet

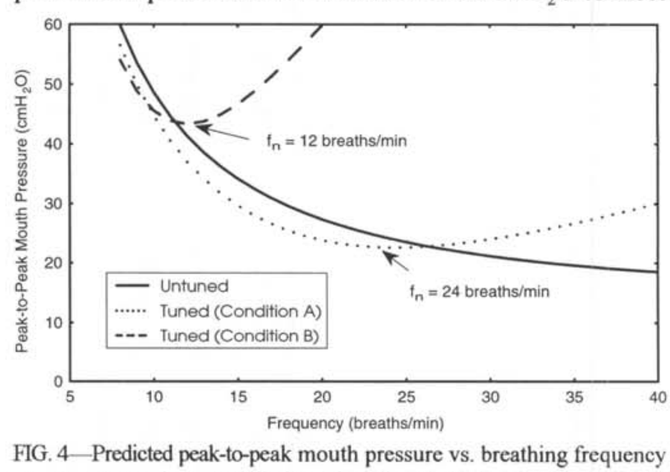

FIG. 4—Predicted peak-to-peak mouth pressure vs. breathing frequency at a constant minute ventilation of 46 liters/min for the closed-circuit system tuned to conditions A and B. *Arrows* show the resonant frequencies (f_n) for each condition. Curve shown by the *solid line* represents the closed-circuit system characteristics without the benefit of tuning (i.e., with minimal inertance in the system) at the same constant minute ventilation. Untuned case does not have a f_n. All curves were derived from the mathematical model described (Equation 7) by Joye and Wechgelear (2), using the values of system inertance, elastance, and resistance given in the legend to Fig. 1.

ventilatory needs. By reducing their f_b closer to the f_n [i.e., 0.20 Hz (12 breaths/min)] and increasing their V_T they were able to maintain their ventilatory requirements with substantially lower peak-to-peak mouth pressures. In contrast, the f_n for condition A was very close to the group mean open-circuit f_b during exercise; thus most subjects did not have to change their breathing pattern appreciably to minimize their peak-to-peak mouth pressures when breathing on the system under condition A.

Figure 4 also shows the potential benefit of reactance tuning. When the untuned system is compared with the system tuned to condition A it can be seen that for breathing frequencies below 26 breaths/min, lower ΔPm_{pp} are required to maintain a given ventilation on the tuned system. However, during exercise, unless subjects are able to breathe at frequencies less than 12 breaths/min, no benefit is gained from reactance tuning when the system is tuned to condition B.

As discussed in Joye and Wechgelaer (2), the amount of flow resistance added by including a tuning device in a closed-circuit system dictates whether a significant benefit can be gained from reactance tuning. If the added resistance is too great, the benefits of tuning are lost. Furthermore, as f_b increases, flow resistance tends to become a more dominant source of impedance in the closed-circuit, thus reducing the relative benefits of reactance tuning at high ventilatory rates. Figure 4 shows that with the current UBA simulator design there is no benefit to be gained from reactance tuning at f_b's greater than 25 breaths/min.

The changes in breathing pattern observed during exercise under the different conditions had little effect on PET_{CO_2} until the f_n of the system was set at the lowest frequency (i.e., 0.20 Hz, 12 breaths/min). Under these latter conditions PET_{CO_2} was significantly higher than during open-circuit breathing, indicating a slight degree of CO_2 retention. Even though the group mean and maximum individual increase in PET_{CO_2} were only 2 and 4 mmHg, respectively, chemical feedback resulting from these small increases in PCO_2 may have played a significant role in the load-compensating responses to the increases in breathing impedance (6).

It is important to note that the respiratory responses of divers at surface are different from their responses at increased ambient pressure (7). It is well known that exposure to a raised PO_2 and increased gas density results in hypoventilation and hypercapnia during exercise at depth (8). Particularly marked CO_2 retention has been observed in divers who exhibit a distinctive slow and deep-breathing pattern during exercise at depth (9,10).

Since changes in gas density will also alter the inertial and resistive components of impedance in UBA, the diver–UBA interactions observed in the present study at surface pressure may not be the same at depth. Clearly the next step in the current research would be to repeat the experiments at an increased ambient pressure or with a raised gas density to understand diver responses to a tunable UBA under more operational conditions.

Respiratory sensation and control of breathing pattern

Mead (11) theorized that for a given level of alveolar ventilation there is a particular respiratory frequency that is least costly in terms of the average force of the respiratory muscles. His findings showed that man at rest breathes at frequencies that are very close to this theoretical resonant frequency. In the present study, when the subject breathes on the tunable close-circuit system, the resonant frequency of the system as a whole is dependent on both the divers lung characteristics and the characteristics of the closed-circuit breathing system.

Our findings indicated that when the f_n of the system as a whole is altered by changing the inertance of the external breathing circuit, subjects instinctively deviate from their normal breathing frequency toward the new resonant frequency of the system. Since the peak-to-peak pressure is directly proportional to the average force of the respiratory muscles during a given respiratory cycle, these results support the idea that at a given minute ventilation f_b is regulated so as to minimize the average force of the respiratory muscles (11).

However, the precise neural mechanisms by which our subjects changed their breathing pattern in response to the different respiratory impedances is uncertain. This is because the mechanisms mediating behavioral respiratory load compensation are still not fully understood. Chonan et al. (12) have suggested that the optimization of breathing may involve minimizing the sensations of respiratory effort and discomfort. Neural input from pulmonary afferents in the airways, lungs, or pulmonary circulation are thought to provide information on the quality of the sensation during loaded and unloaded breathing. This afferent information may then be used to modulate the central command signal to the respiratory muscles to optimize the pattern of breathing (13). The sense of respiratory effort is thought to be mediated through an awareness of the centrally generated respiratory motor command signal by means of collateral discharge within the central nervous system (14).

When breathing on the simulated closed-circuit system, the peak tension generated by the respiratory muscles will be least when the respiratory frequency coincides with the system resonant frequency (Fig. 4). Since respiratory effort should also be minimized when breathing at the system f_n, it would seem appropriate that under the present conditions the sense of respiratory effort be the dominant mechanism for regulating breathing pattern. Indeed, the subjective comments of the divers suggest that they *consciously* changed their breathing pattern to improve breathing comfort when exposed to the different experimental conditions. This observation does not discount the possibility that breathing may have also been altered automatically by the increases in respiratory discomfort and by reflex mechanisms.

A voluntary change in the breathing pattern away from the spontaneous adopted level has also been shown to intensify dyspnea sensations (12). In view of this, we thought it possible that changing the breathing pattern to decrease respiratory effort may at the same time increase feelings of dyspnea. Evidence showing that dyspnea associated with hypercapnia is dissociated from the sensation of respiratory effort has recently been provided by Demediuk et al. (4). It was for this reason that we asked the subjects to separately identify and rate sensations of RE and BC.

The observed changes in breathing pattern away from the spontaneous (open-circuit) frequency toward the f_n of the system suggests that the ameliorating effects of changing breathing pattern to reduce respiratory driving pressures (and hence the sense of respiratory effort) outweigh any dyspnea sensations that may accompany the changes in V_T and f_b per se. It should be noted that the changes in breathing pattern had minimal effects on $P_{ET_{CO_2}}$ and therefore it is unlikely that sensations of breathing discomfort were significantly affected by hypercapnia. The similar pattern of results for RE and BC suggest that under the current experimental conditions these sensations were related in reflecting the perceived level of impedance in the breathing circuit.

Learning effects and repeatability of responses while breathing on the tunable closed-circuit system

As none of the subjects had previously breathed on a tunable closed-circuit system, it was anticipated that there may be a training or learning effect as they discovered that while breathing on the system they may improve the mechanical efficiency of their breathing by altering their natural breathing pattern. If subjects changed their normal f_b to one that more closely matched the f_n of the system they could reduce their respiratory driving pressures and hence lower their respiratory effort.

The results showed some evidence of a learning effect in that f_b at rest more closely followed the changes in the f_n of the system during the repeat trials than during the initial trials. It is possible that this change in f_b may have been partly responsible for the lower ratings of RE and improved BC observed at rest during the repeat experiments. However, the changes in resting f_b resulted in respiratory driving pressures that were only marginally lower and not significantly different from those during the initial trials. Therefore, it seems likely that other factors, possibly behavioral in origin, could have contributed to the lower ratings at rest during the repeat experiments.

Wilson and Jones (15) have shown that breathlessness sensations during exercise (defined as "an uncomfortable need to breathe") are highly reproducible over multiple experiments carried out over several weeks when rated using the Borg scale. However, in a separate study (16), the same authors have reported significantly lower ratings of breathlessness using both the Borg and a visual analog scale during a repeat exercise trial conducted 2–6 wk after the initial trial. These authors suggested that the lower subjective ratings given during the repeat trial could be due to a change in the subject's perception of breathlessness (16).

In the present study a change in the subject's perception of RE and BC during the repeat trial could possibly have resulted from an increased familiarity with the respiratory sensations associated with breathing on the tunable closed-circuit system. Despite minimal changes in respiratory driving pressures between the initial and repeat trials, the reduced novelty of the respiratory sensations during the repeat trial may have contributed to a lower perception of respiratory effort and improved sense of BC.

During exercise there were no changes in either respiratory responses or ratings of RE and BC between the initial and repeat tests. This suggests that familiarization with breathing on the apparatus seems to have occurred quickly (i.e., within the initial trials at rest). Freedman and Weinstein (17) have shown that repeated presentation of elastic loads leads to a change in the response to loaded breathing so that ventilation is restored more quickly during subsequent exposures. The exact nature of the newly adopted pattern of breathing depends on the size and the physical characteristics of the load and is thought to be a learned response mediated by voluntary mechanisms (17).

Comparison of empirical findings with model predictions

The respiratory pressures generated while breathing on the tunable simulated closed-circuit UBA are dependent on the interaction between the individual's choice of breathing pattern and the physical characteristics of the system. The physical characteristics of the simulated closed-circuit UBA are governed by its design and geometry and can be modeled mathematically to predict its burden on the diver's respiratory system (i.e., the external respiratory load).

Using Equation 8 from Joye and Wechgelaer (2), we predicted the ΔPm_{pp} that would be generated as a function of breathing frequency for V_T isopleths between 1 and 3 liters using the values of system resistance, inertance, and elastance given in the legend of Fig. 1.

Figure 5 shows the results of the model predictions when the system is configured to conditions A and B. In each plot on Fig. 5 are data points showing model predictions for ΔPm_{pp} using the group mean data for V_T and f_b obtained while breathing on the closed-circuit at rest and while exercising (Tables 4 and 5). The error bars show one standard deviation either side of the mean for f_b and ΔPm_{pp}. The ΔPm_{pp} predicted by the mathematical model agreed closely with our observations. For condition A the average ($\pm$ SD) ΔPm_{pp} was 22.7 ± 4.54 cmH_2O at a f_b of 21.9 breaths/min and V_T of 2.17 liters. The model predicted a ΔPm_{pp} of 23.7 cmH_2O, only marginally higher but within 1 SD of the mean of the observed value. At rest, the average ($\pm$ SD) ΔPm_{pp} was 12.9 ± 3.03 cmH_2O at a f_b of 14.8 breaths/min and a V_T of 1.24 liters. The model predicted

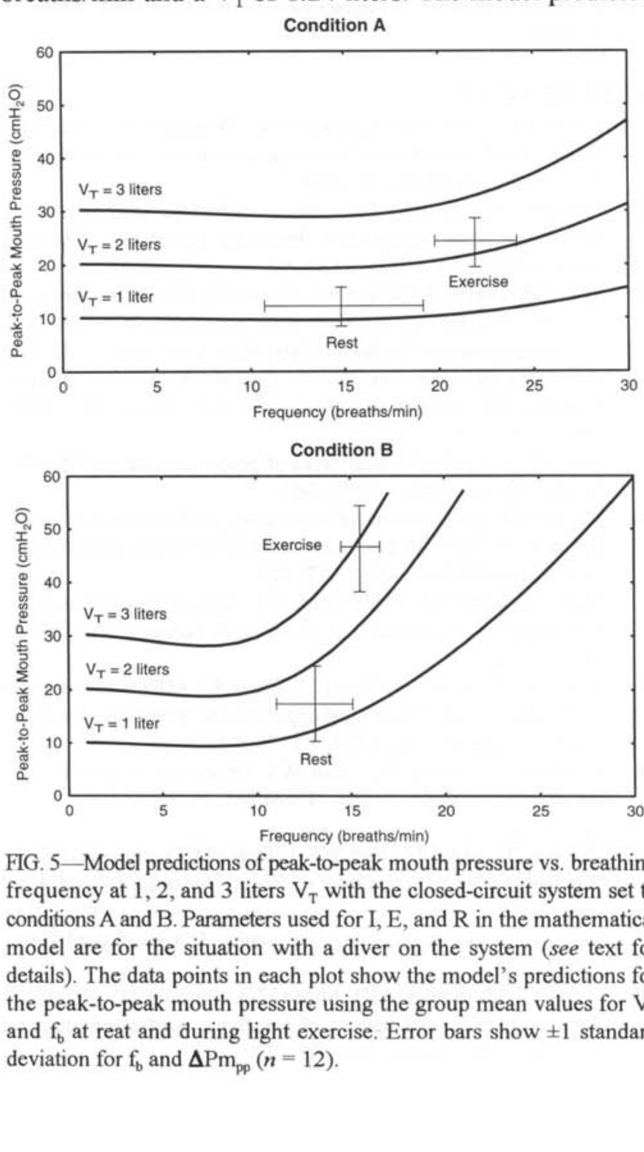

FIG. 5—Model predictions of peak-to-peak mouth pressure vs. breathing frequency at 1, 2, and 3 liters V_T with the closed-circuit system set to conditions A and B. Parameters used for I, E, and R in the mathematical model are for the situation with a diver on the system (*see* text for details). The data points in each plot show the model's predictions for the peak-to-peak mouth pressure using the group mean values for V_T and f_b at reat and during light exercise. Error bars show ± 1 standard deviation for f_b and ΔPm_{pp} (n = 12).

11.9 cmH_2O, a value slightly lower, but again well within 1 SD of the mean.

Similar results are shown for the valve closed position (condition B). During exercise, the average ($\pm$SD) ΔPm_{pp} was 46.1 ± 8.46 cmH_2O at a f_b of 15.5 breaths/min and a V_T of 2.90 l. The model predicted a ΔPm_{pp} of 46.5 cmH_2O. At rest, the average ($\pm$SD) ΔPm_{pp} was 17.7 ± 7.06 cmH_2O at a f_b of 13.1 breaths/min and a V_T of 1.39 liters. Here the model predicted a ΔPm_{pp} of 17.4 cmH_2O. The above comparisons show the predictive capability of the model to be excellent under these conditions.

The model predictions in Fig. 4 show the resonant frequencies for conditions A and B at 24 and 12 breaths/min, respectively, whereas in Fig. 5 the points of minimum ΔPm_{pp} are somewhat lower than the resonant frequencies. As discussed in Joye and Wechgelaer (2), this downward shift in frequency of the point of minimum pressure from the undamped resonant frequency is a result of the contribution of resistance relative to the inertance within the tuning circuit when the f_n of the system is determined at a constant V_T.

Using Equation 11 in Joye and Wechgelaer (2), the minimum ΔPm_{pp}, which reflects the damped f_n of the closed circuit, was calculated to occur at frequencies of 13.3 and 7.3 breaths/min for conditions A and B, respectively. However, when $\dot{V}_E$ rather than V_T is held constant the f_n of the system becomes independent of the system's resistance and there is no shift in the minimum pressure point under these conditions (2). Since our subjects showed relatively little change in $\dot{V}_E$ during exposure to the different closed-circuit conditions we felt it was more appropriate to compare breathing responses with the undamped rather than the damped f_n.

The breathing frequency at which ΔPm_{pp} will be minimized for the system as a whole (i.e., diver plus UBA) will vary slightly between divers due to individual differences in total lung volume. During tidal breathing, system volume will fluctuate; however, these lung volume changes have only small effects on total system elastance. Although lung volume was accounted for in our model (because it has an effect on the water column elastance) the effects of chest and lung wall elastance was not. We chose not to include respiratory elastance in our model for several reasons. First, elastance of the respiratory system has no effect on mouth pressure, which we used as our driving force for the model. Second, modeling the respiratory system with an electric circuit analog is not straightforward. If the combination of respiratory elastance and UBA elastance reduces the overall system elastance by half, then electric circuit analog theory would predict that the resonant frequency of the system would change by 30% (1/√2). However, if the lung and chest wall elastance is modeled as a shunting capacitance, the resonant frequency of the system as a whole is not affected (J.R. Clarke personal communication,).

Determining an appropriate electric circuit analog for the inclusion of the lung and chest wall elastance in our model is complicated because the respiratory system is the driver (i.e., power source) for our system. Consequently, the combinations of physical elastances (i.e., respiratory elastance and UBA elastance) does not easily correspond with electric circuit analogs, thus making the issue of the effects of respiratory elastance on the system's resonant frequency a much more complicated matter than can be determined from simplified linear electric circuit models.

Attempts to model the effects of lung and chest wall elastance are also confounded because elastance of the chest and lung is nonlinear and is a function of V_T and end-expiratory lung volume, both of which change with exercise (18). Values for respiratory elastance also vary dependent on whether it is measured on inspiration or expiration. Furthermore, respiratory elastance determined from static relaxation pressure-volume curves differs from the dynamic elastance measured during breathing (18). It is clear however, that whatever the effects of respiratory elastance are, it does not noticeably shift the resonant frequency of the system as a whole, since the actual data are predicted reasonably well by our model.

SUMMARY AND CONCLUSIONS

Clearly, the respiratory pressures generated under the closed-circuit conditions, especially during exercise, are greater than would be acceptable in a real closed-circuit UBA. However, the main objective of the current UBA simulator design was to emphasize the effects of reactance tuning by introducing noticeable changes in the system's characteristics within the expected physiologic range of breathing frequencies. In this manner we were able to study how changes in the f_n of a simulated closed-circuit UBA influence breathing pattern and respiratory comfort. The main conclusions emanating from this study are summarized below.

1. When the simulated UBA was switched from open- to closed-circuit breathing subjects changed their f_b in a direction toward the f_n of the system and attempted to maintain minute ventilation constant by adjusting tidal volume.
2. Behavioral reactions aiming to minimize the sense of respiratory effort or breathing comfort or both offer a possible mechanism by which breathing pattern is regulated in response to changes in the f_n characteristics of a tunable closed-circuit UBA.
3. Empirical data for mouth pressure recorded at rest and during light exercise agrees well with model predictions of system impedance for the simulated closed-circuit UBA tuned to various resonant frequencies.

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The views expressed herein are strictly those of the authors and are not to be construed as reflecting those of the U.S. Navy, Department of Defense, or the U.S. Government.

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