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Virtual Patient Simulator for the Perfusion Resource Management Drill

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Abstract: Perfusionists require a detailed understanding of a patient's physiological status while comprehending the mechanics and engineering of the cardiopulmonary bypass system, so it is beneficial for them to obtain relevant practical skills using extracorporeal circulation technology and educational physiological simulators. We designed a perfusion simulator system (ECCSIM: Extracorporeal Circulation SIMulator system) based on a hybrid of a simple hydraulic mock circulation loop linked to a computer simulation model. Patient physiological conditions (height, weight, and cardiac indices) were determined by a parameter estimation procedure and used to accurately reproduce hemodynamic conditions. Extracorporeal circulation trainees in pre-clinical education were able to maintain venous oxygen saturation levels above 50%, except during cardiac standstill and a brief resumption of pulsation. Infant amplitudes of reservoir volume oscillation and flow rate were greatly increased compared with adult cardiovascular parameters, this enabled the instructor to control the difficulty level of the operation using different hemodynamic variations. High-fidelity simulator systems with controllable difficulty levels and high physiological reproducibility are useful in constructing a perfusion resource management environment that enable basic training and periodic crisis management drills to be performed.
Keywords: cardiopulmonary bypass, computer simulation, hemodynamics, perfusion, physiology. *JECT. 2009;41:206–212*

During cardiopulmonary bypass, human error can be fatal. Since perfusionists require details about the patient's physiological status at the same time as an understanding of the system mechanics and engineering, it is useful to develop physiological simulator systems that provide physical feedback for the practical skill acquisition of extracorporeal circulation technology.

The Extracorporeal Circulation Simulator system (ECCSIM and ECCSIM-Lite), an educational simulator for the acquisition of such skills, has been under continuous development since 2003 (1–4). The system was designed as an application of a hybrid (numerical-hydraulic) model (5) that reproduces patient hemodynamics using minimum components of constitution.

In this paper, we describe the advantages and application of a patient simulator system that builds a physical interface as the institutional environment and that reproduces the patient's physiological condition using a simple hydraulic mock circulation loop linked to a computer simulation model. Blood vessel circuit parameters obtained from various patient physiques (4) were adapted to extracorporeal circulation training for pre-clinical residents. The system was confirmed as an adequate core element for Perfusion Resource Management (PRM) education.

MATERIALS AND METHODS

System Description

The system is composed of a simple hydraulic mock circulation loop (hydraulic section) and a personal computer containing the numerical simulation model (virtual patient model) that simulates vital patient behaviors and controls the hydraulic section (Figure 1). The mock circulation loop continuously feeds back physiological information such as arterial line pressure, arterial, venous, vent and suction flow, and blood volume to the numerical simulation model.

The hydraulic section is connected to the institutional heart lung machine, and the trainee can operate

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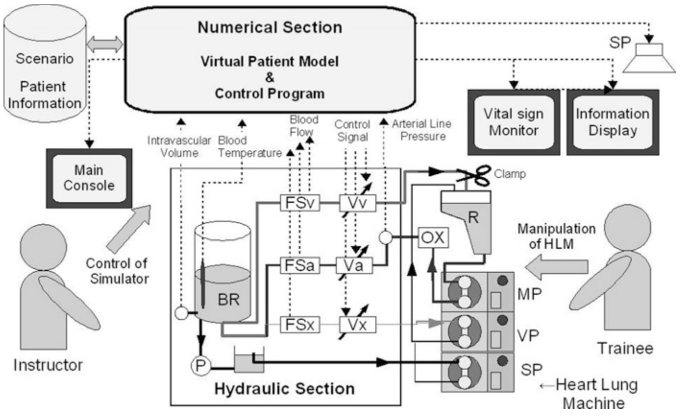

Figure 1. Schematic block diagram of numerical-hydraulic simulator system. DB: Database of predefined scenario and operation record; FSa, FSv, FSx: Flow sensor unit of arterial, venous, and additional port respectively; BR: Blood capacitor reservoir; Va, Vv, Vx: Flow control valve unit of arterial, venous, and additional line respectively; MP: Main pump unit; VP: Vent or suction pump unit; SP: Suction pump unit.

the same equipment as that used in the operating theater while monitoring vital signs (Figure 2). Since the proportional controlled valve is included in each line of circuit, arterial, venous, and vent line, flow can be controlled using arbitrary timing from the instructor side without the trainee noticing. The proportional controlled valve can also be used to adjust arterial line pressure and venous flow.

The vital sign monitor displays ECG and blood pressure readings to the trainee (Figure 3A), while the operating information monitor displays arterial flow, vent flow, arterial pressure, oxygen flow, and blood-gas information (Figure 3B). The trainee can infuse drugs and measure the activated coagulation time (ACT) using touch-panel operation of the operating information monitor. The instructor can control events during operations such as aorta clamping and defibrillation by conveying information on a patient's condition and the heart lung machine status from the supervisor interface (Figure 4).

Patient hemodynamic models can be created from the fluid mechanics element (6), but replicating patient variation becomes very complex. A simple lumped parameter model that presumes component parameters is therefore preferable. Cevenini et al. previously duplicated fundamental hemodynamic behavior using the closed systemic-circulatory loop which consists of only three elements, and were readily able to estimate each parameter (7). Ferrari et al. previously built a computer simulation system to comprehend patient hemodynamics and physiology, enabling the duplication of patient status variation using a combination of different component modules (8).

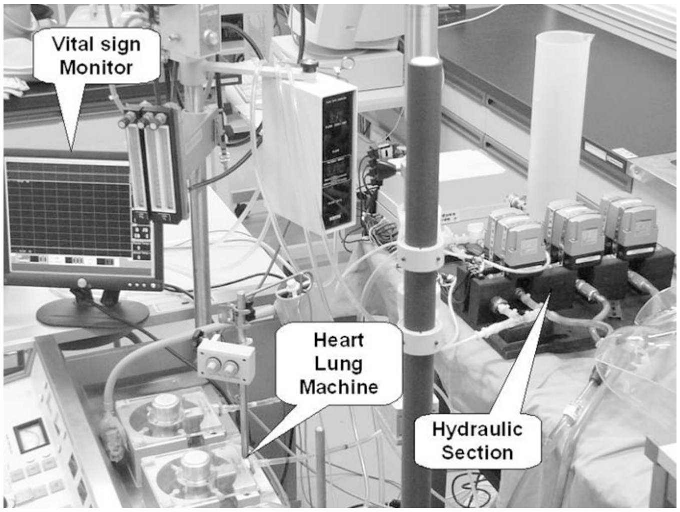

Figure 2. Hydraulic section of simulator connected to heart lung machine. Hydraulic analog is positioned on an operating table and physically connected to circuit of heart lung machine.

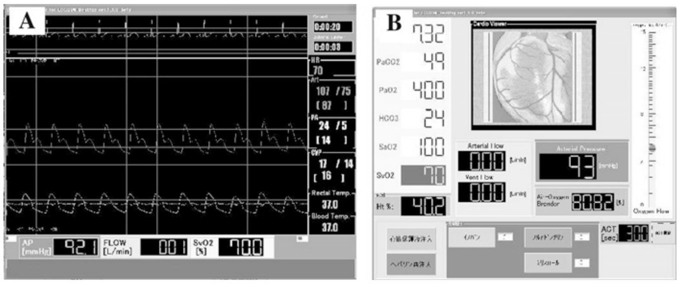

Figure 3. A, Visual user interface for trainee. Vital sign monitor panel; B, Information panel of blood gases, arterial pressure, and line flows of each circuit line.

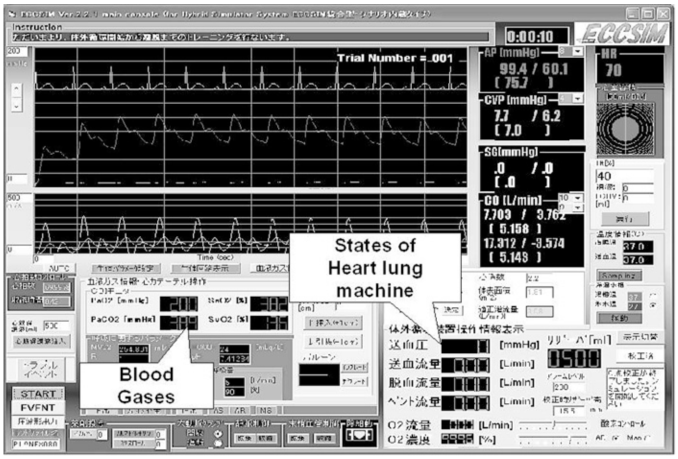

Figure 4. Visual user interface for instructor. Interface is made up of vital sign monitor panel, blood gases monitor panel, status display of heart lung machine, and other controls for instructor's operation.

In the present system, patient hemodynamics with extracorporeal circulation were reproduced using numerical simulation in which the arterial line port, venous line port, and vent port were connected to the closed loop of the systemic circulation containing the heart and pulmonary circulation (Figure 5). The lumped parameter model was used with reference to Ferrari et al.'s model (8). The modified windkessel model was adopted as the arterial vessel system. The simple windkessel model was adopted as the venous vessel system to duplicate the arterial-blood-pressure wave with the minimum components. The variable compliance model was used as the kinetics of the right and the left heart. The flow of each resistance, reactance and capacitance components and nodal point pressures were solved by lower and upper triangular matrices decomposition to solve systems of linear equations quickly.

Microsoft Visual Basic 6.0 was used as the programming language. An IBM-compatible PC (OS: Windows XP SP3, Microsoft Corporation, Redmond, WA, CPU: Pentium Core2-Duo, Santa Clara, CA, 1.2 GHz, Memory: 4GB) was used for simulation. The computation time of one step was less than 0.02 seconds. Hence the trainee was provided with real-time information of a patient's vital signs with sufficient time resolution.

The heart function and arterial system were given as a constant (Table 1). The blood volume (BV), venous compliance (C_{SV}), pulmonary vein compliance (C_{PV}), peripheral resistance, and unstressed volume (V_{min}) were estimated from patient information since these parameters greatly contribute to patient hemodynamics. According to the findings of Cavalcanti and DiMarco (9) and Shoukas and Sagawa (10), BV, C_{SV}, and C_{PV} are directly computed from the weight (W). The peripheral resistance was estimated from the cardiac output (CO) by the following process (Appendix 1). The simulation analysis on these models was carried out systematically in a range of cardiac indices (CI) = 2.0–4.0, weight (W) = 10–120 kg. The relationship of initial peripheral resistance (R_{m0}) and CO was derived for maintaining the mean arterial pressure (AP) at 85 mmHg. The estimation procedure of the unstressed volume was determined by multiple regression analysis of CI, and weight change on AP was 85 mmHg (Appendix 2).

Peripheral resistance and unstressed volume were determined as follows (Figure 6). CO was calculated from patient physical information (height, weight, body-surface area, and CI), R_{m0} was determined from CO, and peripheral resistance was derived by the change of AP.

Blood gas information was determined using the formula displayed in Figure 7. The efficiency of oxygen transfer in the artificial lung K, the respiratory quotient RR, bicarbonate consistency $[HCO_3]$ (mEq/L), hemoglobin concentration tHb (g/dL), and patient weight (kg) were given as constants. The gas flow V_L (mL/min) and the oxygen consistency FiO_2 were determined by trainee knob manipulation. The cardiac output CO (mL/min) was determined as flow of the systemic-circulation loop calculated by the detection of the arterial line flow Q_{ALF} (mL/min), the venous line flow Q_{VD} (mL/min), and unstressed volume V_{min}(mL). The oxygen consumption VO_2 (mL/min)

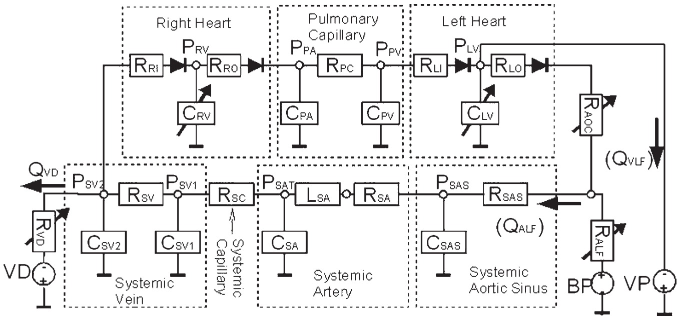

Figure 5. Electrical analog of systemic circulatory system including extracorporeal circulation. The systemic circulatory loop is composed of a modified or simple windkessel model and the variable compliance model. The main pump (BP), venous drainage line (VD), and vent pump (VP) are connected to the systemic circulatory loop.

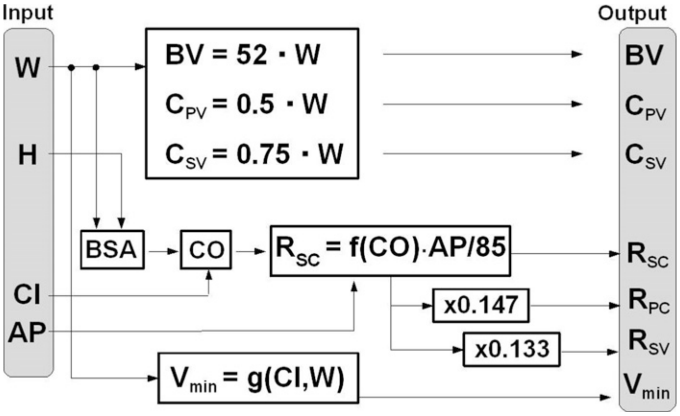

Figure 6. Schematic representations of cardiovascular parameter estimation procedure. Blood volume, venous compliance, and pulmonary vein compliance are computed from body weight directly. Cardiac output is computed from body surface area (BSA) computed from W and height (H), then systemic peripheral resistance is derived by presumption model f(CO) shown in Appendix 1. Relationship of unstressed volume and W and CI is then derived. Unstressed volume is derived by presumption model g(CI, W) shown in Appendix 2. The oxygen consumption volume is derived by presumption model h(W, Temp) shown in Appendix 3.

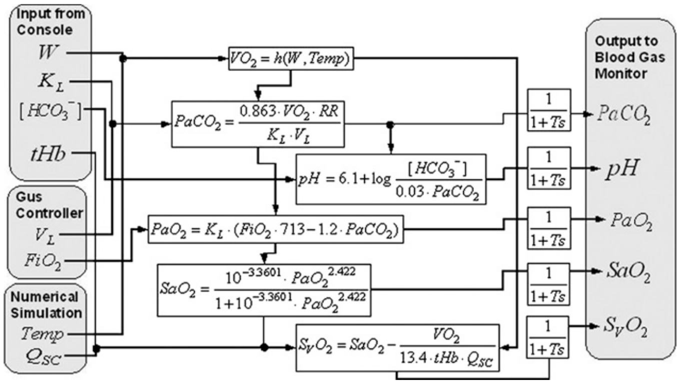

Figure 7. Schematic representations of blood gas models. The model is made up of three input compartments (from instructor's console, gas controller for trainee, and simulated results of numerical section), several experimental or analytical equations, and delay for blood gas monitor display. T is time constant of delay set as 2.0 seconds.

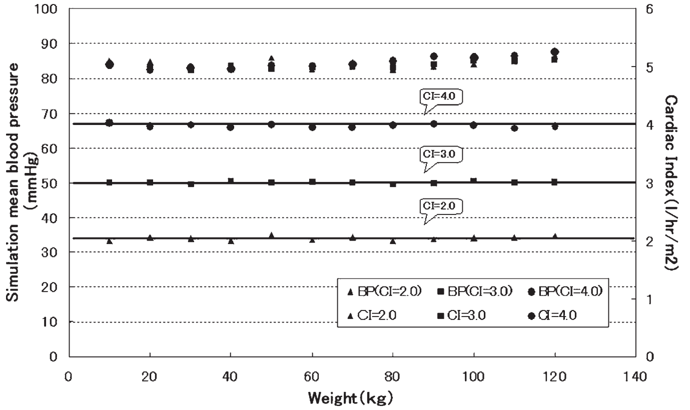

Figure 8. Comparison of target and simulated cardiac index and mean blood pressure to body weight. Simulated mean blood pressure was maintained at target value (85 mmHg), and simulated cardiac index is according to target value (2.0, 3.0, and 4.0) in a range of 10–120 kg body weight.

was estimated from the body weight and body temperature (Appendix 3). Arterial oxygen pressure PaO_2, venous blood oxygen pressure $PaCO_2$, arterial oxygen saturation SaO_2, venous blood oxygen saturation SvO_2, and pH were derived from typical blood gases scenarios (11).

Drill Parameters

The validity of the cardiovascular parameter estimation procedure was examined in the weight range 10–120 kg by each simulation when the cardiac index CO was 2.0, 3.0, and 4.0. To investigate how the difference in patient parameters affected operation difficulty, both extracorporeal adult and infant circulation training procedures (H = 1.67 m, W = 60 kg, and H = .8 m, W = 10 kg, respectively) were carried out.

The subject was an undergraduate student from the Department of Clinical Engineering, Hiroshima International University, Japan who attended the pre-clinical education lecture course and had no clinical experience of cardiopulmonary bypass. One of the authors acted as instructor according to the standardized cardiopulmonary bypass procedure. All physiological data during the operation during training (various line flows, reservoir volume, and so on) and vital signs (blood pressure, flow, blood gas values, and so on) were continuously recorded.

RESULTS

Validation of Cardiovascular Parameters

The simulation analysis was derived for a range of weights (10–120 kg), the target of mean AP was 85 mmHg and the CI ranged from 2.0–4.0. We confirmed that the target of CI was in accordance with simulated CI from CO (Figure 8).

The estimated infant cardiovascular parameters (weight 10 kg, height .8 m) were compared with the clinical value of Goodwin et al.'s study (12). According to these results, the V_{min} was 130 mL, the systemic peripheral vessel resistance (Rsc) was 2.00 mmHg·mL^{-1}·sec, the systemic vein compliance (Csv) was 10.0 mmHg/mL and the systemic vein resistance (Rsv) was .18 mmHg·mL^{-1}·s. In our estimation, V_{min} was 161 mL, Rsc was 3.02 mmHg·mL^{-1}·sec, Csv was 10.0 mmHg/mL, and Rsv was .42 mmHg·mL^{-1}·sec. The present estimation procedure therefore produced similar results to clinical settings. It would be expected that the fixed heart rate of 70 bpm influences the higher estimated Rsc and Rsv results than those observed by Goodwin et al.

Simulation Drill

Hemodynamic (Figure 9A), blood-gases (Figure 9B), and pump operation information (Figure 9C) were obtained during the adult extracorporeal circulation training session. Venous oxygen saturation was maintained above 50% except for cardiac standstill and the short period of pulsation resumption.

The trainee was instructed to check a variation in blood-gases information while altering the artificial-lung oxygen flow. The trainee also underwent an infant extracorporeal circulation training session (Figure 10), during which, the amplitude of reservoir volume and flow rate increased more than it had for the adult case. As a result, the operation increased in difficulty and the trainee could not maintain venous oxygen saturation above 50% for about 1 minute after cardiac standstill.

Table 1. Cardiovascular Parameters for Systemic Circulatory System Electrical Analog.

Systemic Resistance (mmHg·mL^{-1}·s)	Systemic Compliance (mL·mmHg^{-1})	Systemic Reactance (mmHg·s^{2}mL^{-1})
Aortic R_{SAS} = .01	Right ventricular C_{LV} = .1–30	Artery L_{SAT} = .01
Arterial R_{SA} = .05	Left ventricular C_{RV} = 1.8–15	
Aortic value R_{LO} = .01–1000	Aortic C_{SAS} = 1.6	
Mitral valve R_{LI} = .005–1000	Arterial C_{SA} = .4	
Pulmonary valve R_{RO} = .01–1000	Pulmonary artery C_{PA} = 1.5	
Tricuspid valve R_{RI} = .005–1000	Venous C_{SV1} = C_{SV2} = W × .5	
Pulmonary R_{PC} = R_{SC} × .147	Pulmonary vein C_{PV} = W × .75	
Venous R_{SV} = R_{SC} × .133		

Weight, W (kg); Blood volume, V_0 = W × 52 (mL).
Peripheral resistance R_{SC} estimated by cardiac output.

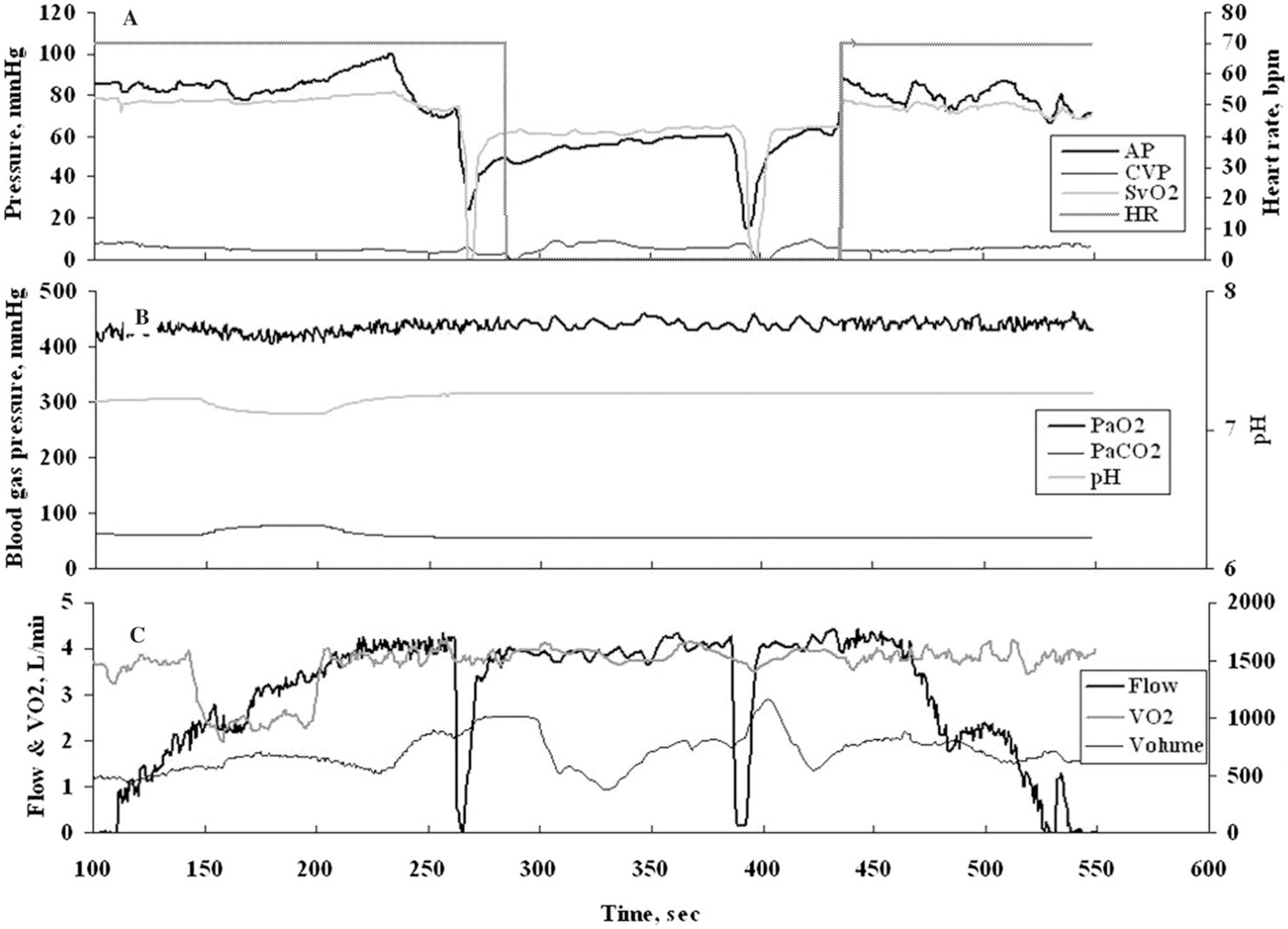

Figure 9. A, Time courses of arterial pressure, centrifugal venous pressure, venous oxygen saturation, and heart rate; B, arterial oxygen pressure, carbon dioxide pressure and pH; C, arterial flow rate, oxygen gas flow rate and reservoir volume. Simulation training is carried out using adult cardiovascular parameters (H = 1.67 m, W = 60 kg).

DISCUSSION

The virtual patient simulator system was developed using a hybrid (numerical-hydraulic) model. It has previously been confirmed that an augmented reality simulator with physical feedback is more effective for acquisition of practical skills than a virtual reality simulator (13). Since the institutional heart-lung machine was physically connected to the present system, the augmented physical feedback interface was realized in this study. The instructor was able to control the difficulty of the training by reproducing hemodynamic variations for different patients from arbitrary patient information.

High-fidelity simulation systems have been developed to effectively reduce human error (14). In the educational field of anesthesia, many patient simulators were developed during the 1960s, and it is recognized that improvements in physiology fidelity levels are important for the teaching of psychomotor skills (15). Morris and Pybus (16) designed an advanced perfusion simulator system containing information on hemodynamics, blood gases, oxygen consumption, and pharmacodynamics for use within a specialized simulation facility; they adopted a hybrid model such as the one used in the present study. Turkmen et al. (17) also chose a hybrid model for a simulator suitable for the education of perfusion students. Since the hybrid model utilizes variation from different patients, one system has the potential for application in both basic training and crisis management drills.

In a numerical simulation of the virtual patient model, variations in conditions can be increased as the number of elements increases. Specialized instructor knowledge is required in such cases as the determination procedure of these parameters becomes more complicated. Bauernschmitt et al. (18) reported that the expression of the cardiovascular system with a multiplex branch mechanism of 128 segments can simulate different extracorporeal circulations. As the systemic circulation of our system

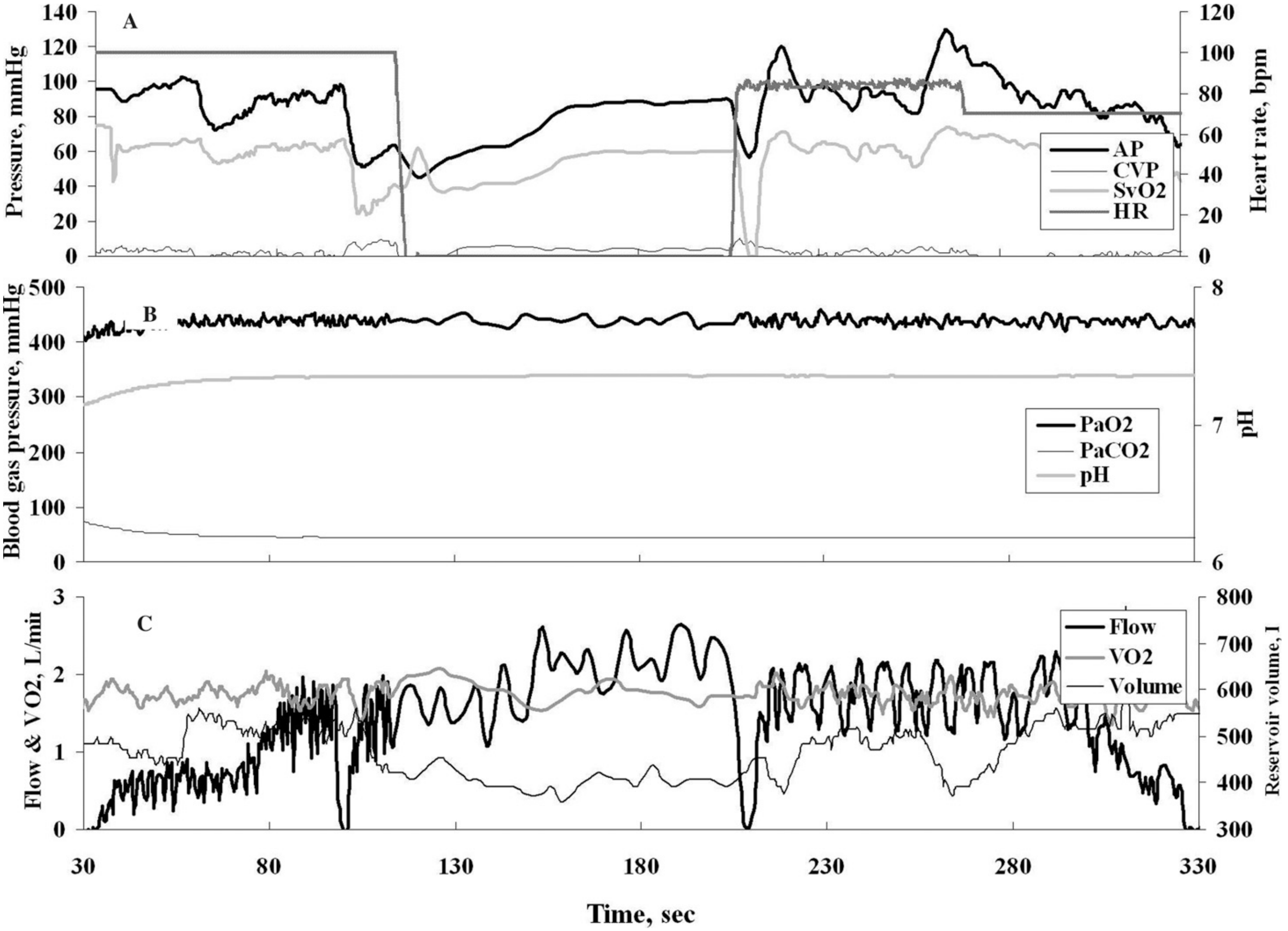

Figure 10. Time courses of vital signs and operating results. Simulation training is carried out using infant cardiovascular parameters (H = 0.6 m, W = 10 kg).

is composed of a minimum of 17 segments, it is recommended that a typical patient's hemodynamics be reproduced directly by input of patient basic information.

The difference between habituation learning from a simulator operation and genuine knowledge of clinical skills is an important shortfall of simulation training. However, since our system enables vital reactions and unexpected scenarios to be practiced by the trainee, it is applicable to evaluation of practical skills that are beyond mere habituation.

In 2006, the number of Japanese certified perfusionists was listed as 569; their nationwide distribution is not uniform, and the per capita average number of operations experienced is less than 100 (1). Moreover, 85% of crisis management drill participants did not carry out institutional periodical trouble management training (19). To shorten the training period and compensate for inexperience, a high-fidelity environment is expected to make Perfusion resource management (PRM) training possible in a shortened training period. Seropian (20) found that a plausible environment, responses and interactions, as well as familiar and realistic simulation equipment are required for a successful simulation.

Our system succeeds in offering a plausible response, patient interaction, and familiar equipments. To realize widespread PRM drills, our future development target lies in the construction of appropriate training scenarios and the quantitative evaluation of skills.

CONCLUSION

In the present study, we developed an extracorporeal circulation simulator system, for use as an educational simulator in the acquisition of basic skills and crisis management in perfusion procedures. This hybrid (numerical-hydraulic) model can reproduce patient hemodynamics using minimal components of constitution.

The instructor can control the level of difficulty of training by using different patient hemodynamic variations from arbitrary patient information. We believe that our system successfully constructs a realistic PRM environment to enhance basic training and periodic crisis management drills to be carried out.

REFERENCES

1. Ninomiya S, Tokumine A, Yasuda T, Tomizawa Y. Development of an educational simulator system, ECCSIM-Lite, for the acquisition of basic perfusion techniques and evaluation. J Artif Organs. 2007;10:201–5.
2. Tokumine A, Ninomiya S, Anno M, et al. Educational simulator "ECCSIM-Lite" used at the JaSECT seminar with analysis of questionnaire survey results Jpn J of Extracorporeal Technology. 2008;35:343–7 (in Japanese).
3. Tokaji M, Ninomiya S, Tokumine A, Kurosaki T. Study of the hemodynamics parameter estimation procedure of the educational perfusion simulator based on basic patient conditions. Jpn J Extracorporeal Technology. 2009;36:24–7 (in Japanese).
4. Ninomiya S, Tokumine A, Tokaji M, Kurosaki T. The extracorporeal circulation simulator System: ECCSIM and ECCSIM-Lite. CIRCULATION Up-to-Date. 2009;4:22–32 (in Japanese).
5. Ferrari G, Kozarski M, De Lazzari C, et al. Modelling of cardiovascular system: Development of a hybrid (numerical-physical) model. Int J Artif Organs. 2003;26:1104–14.
6. Timms D, Hayne M, McNeil K, Galbraith A. A complete mock circulation loop for the evaluation of left, right, and biventricular assist devices. Artif Organs. 2005;29:564–72.
7. Cevenini G, Barbini P, Cappello A, Avanzolini G. Three-element model for total systemic circulation: Emphasis on the accuracy of parameter estimates. J Biomed Eng. 1987;9:374–8.
8. Ferrari G, De Lazzari C, Mimmo R, Tosti G, Ambrosi D. A modular numerical model of the cardiovascular system for studying and training in the field of cardiovascular physiopathology. J Biomed Eng. 1992;14:91–107.
9. Cavalcanti S, Di Marco LY. Numerical simulation of the hemodynamic response to hemodialysis-induced hypovolemia. Artif Organs. 1999;23:1063–73.
10. Shoukas AA, Sagawa K. Total systemic vascular compliance measured as incremental volume-pressure ratio. Circ Res. 1971;28:277–89.
11. Martin L. *All You Really Need to Know to Interpret Arterial Blood Gases 2/E.* Philadelphia: Lippincott Williams & Wilkins Inc; 1999:42–7.
12. Goodwin JA, van Meurs WL, Sa Couto CD, Beneken JE, Graves SA. A model for educational simulation of infant cardiovascular physiology. Anesth Analg. 2004;99:1655–64.
13. Botden SM, Buzink SN, Schijven MP, Jakimowicz JJ. Augmented versus virtual reality laparoscopic simulation: What is the difference? A comparison of the ProMIS augmented reality laparoscopic simulator versus LapSim virtual reality laparoscopic simulator. World J Surg. 2007;31:764–72.
14. Higton P. Safety lessons from aviation. Perfusion. 2005;20:191–3.
15. Cumin D, Merry AF. Simulators for use in anaesthesia. Anaesthesia. 2007;62:151–62.
16. Morris RW, Pybus DA. "Orpheus" cardiopulmonary bypass simulation system. J Extra Corpor Technol. 2007;39:228–33.
17. Turkmen A, Rosinski D, Noyes N. A simulator for perfusion training. Perfusion. 2007;22:397–400.
18. Bauernschmitt R, Naujokat E, Mehmanesh H, et al. Mathematical modelling of extracorporeal circulation: Simulation of different perfusion regimens. Perfusion. 1999;14:321–30.
19. Tomizawa Y. Trouble simulation course for extracorporeal circulation: Experiences from wet lab. Jpn J Artif Organs. 2006;35:223–6.
20. Seropian MA. General concepts in full scale simulation: Getting started. Anesth Analg. 2003;97:1695–705.

APPENDIX 1

The estimation formula of the systemic capillary resistance: R_{m0}(mmHg·mL^{-1}·sec) from cardiac output: CO(L/min)

$$R_{m0} = f(CO) = 4.2442 \cdot CO^{-1.0706}$$

APPENDIX 2

The estimation formula of the unstressed volume: V_{min}(mL) from the body weight: W(kg) and the cardiac index: CI(L·m^2·min^{-1})

$$\begin{aligned} V_{min} = G(CI, W) &= 25.2298 \cdot CI^2 - 0.01739 \cdot W^2 \\ &- 1.95691 \cdot CI \cdot W - 145.666 \cdot CI \\ &+ 45.0998 \cdot W - 20.1374 \end{aligned}$$

APPENDIX 3

The estimation formula of the oxygen consumption rate: VO_2 from the body weight: W(kg) and body temperature: Temp(°C)

$$\begin{aligned} VO_2 = h(W, Temp) &= (-0.00001624 \cdot W^4 + 0.036248 \cdot W^3 \\ &- 0.27653 \cdot W^2 + 10.834 \cdot W) \cdot (-0.00029 \cdot Temp^3 \\ &+ 0.0284 \cdot Temp^2 - 0.8509 \cdot Temp + 8.2832) \end{aligned}$$