Preload control of the increased outflow of a dual pulsatile extracorporeal membrane oxygenator

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Abstract

An improved pulsatile extracorporeal membrane oxygenation (ECMO) device is needed to reduce the high risk of complications associated with existing ECMO devices, due to continuous blood outflow (which reduces blood perfusion) and a complex structure that makes setup and management difficult. This study introduces a new pulsatile ECMO device to maintain sufficient pulsatility (an "energy equivalent pressure increment" [EEPI] of at least 20 %) and simplify the structure of the pulsatile pump by removing artificial valves and complex actuators. The hemodynamic characteristics and pulsatility of the proposed pulsatile ECMO device were evaluated in-vitro using a MOCK system. Although the pulsatile ECMO device has the same dual pumping structure as existing pulsatile ECMOs, the newly applied preload control mechanism increases the outflow of the proposed pulsatile ECMO compared to the previous device.

1. Introduction

Extracorporeal membrane oxygenation (ECMO) is an extracorporeal technique of providing prolonged cardiac and respiratory support to the patient with severe cardiopulmonary diseases [1, 2]. An ECMO device comprises a pump for blood perfusion and membrane oxygenator for gas exchange [1-3]. According to the Extracorporeal Life Support Organization (ELSO), the application of ECMO increased by 20.9 % annually from 2010 to 2020. In 2020, 18260 cases in which ECMO was used due to COVID-19 were reported at 521 centers worldwide [4]. In patients with cardiogenic shock, venoarterial-extracorporeal membrane oxygenation (VA-ECMO), which connects the ECMO device from a vein to an artery, is performed, but the survival rate is only 59-72 %, and the risk of complications such as left ventricle dilation, pulmonary edema, and gastrointestinal bleeding is high [3-7]. In-vivo experiments showed that the continuous blood pump incorporated into existing ECMOs is associated with reduced blood flow to the brain, liver, and heart [8, 9]. Therefore, it is necessary to develop a pulsatile ECMO to reduce the risk of complications [4, 10, 11].

When an ECMO device is applied together with an intra-aortic balloon pump (IABP) to supply pulsatile blood flow, the risks of left ventricular (LV) dilation, gastrointestinal bleeding, and mortality are reduced [12, 13]. However, simultaneous application of both devices increases the difficulty of risk management [14]. Although a pulsatile ECMO device has been developed for clinical practice, the pulsatile pump in this system uses multiple valves, making the system complex, expensive, and difficult to apply clinically [4, 10, 14, 15]. Therefore, there have been attempts to develop a pulsatile ECMO device using a simple inexpensive centrifugal pump. An ECMO using a pulsatile centrifugal pump generated pulsatile blood flow by periodically increasing the speed of its impeller. Compared to continuous operation mode, hemolysis occurs more often due to the high maximum rotation speed of the impeller [10, 16]. Because hemolysis is affected by the device's internal structure and material properties, a pulsatile ECMO device was developed with a centrifugal pump that causes less hemolysis and has been applied in animal and clinical experiments [3]. However, the ECMO device using a pulsatile centrifugal pump had similar mortality...
and complication rates to those of conventional ECMO devices, presumably because the pulsatility of the pulsatile centrifugal pump was not sufficient [17].

Undar et al. insisted that the pulsatility of blood flow should be evaluated based on the “energy equivalent pressure increment” (EEPI). Pulsatile ECMO device has 10-20% EEPI, while a pulsatile centrifugal pump has only 0-5% EEPI [18, 19]. Our mobile ECMO device system has a high-pressure tank to supply oxygen during patient transport. However, existing ECMOs are large and heavy, making them difficult to use without a special intensive care facility [20].

Using the pressure of the oxygen tank and several simple solenoid valves, a pulsatile ECMO device can supply sufficient pulsatile blood flow without a complicated motor or actuators. Using the pressure of an oxygen tank and a few simple solenoid valves solves existing problems, and the pulsatile ECMO device can provide sufficient pulsatile blood flow without a complicated motor or actuators. By blocking or opening the tube through which blood flows via oxygen pressure, instead of the input and output valves used in existing pulsatile ECMO device, it is possible to reduce the risk of thrombosis caused by the valves. Like existing pulsatile ECMO device, two pulsatile pumps were used to increase the blood inflow from the vein to the ECMO device [21]. Additionally, by using a preload control device, the stroke volume of the ECMO device was improved compared to existing pulsatile ECMO device, without increasing the priming volume. Since the oxygen tank is a high-pressure environment, even with low power, it can have sufficient pulsatility, like an existing pulsatile ECMO device with 20% EEPI [22, 23].

This study proposed a mobile ECMO device with a simple structure driven by oxygen pressure, and evaluated whether it can supply sufficient blood flow for a patient and has sufficient pulsatility in vitro experiment.

2. Materials and method

2.1 Structure of a pulsatile ECMO device

The developed pulsatile pump ECMO device consists of an oxygen tank, control unit, solenoid valve, air pump, dual pulsatile pump, and oxygenator, as shown in Fig. 1. For the oxygenator, a commercial product (PLS-I, Maquet, Germany) was used, while we manufactured the other components. In the dual pulsatile pump shown in Figs. 1 and 2, the two pulsatile blood pumps consist of a blood inlet/outlet switching blood sac and central blood sac in parallel. The blood inlet/outlet switching blood sac are located on both sides of the central blood sac, and simultaneously suck and discharge blood and act as a valve. The blood sac is driven by the high-pressure oxygen in
the tank and contracts sequentially without a mechanical valve, to discharge blood in one direction. Each blood pump consists of a blood sac made of flexible, blood-compatible polyurethane inside an external acrylic case. The dual pulsatile pumps are connected in parallel and operate at intersections. By sucking and ejecting blood, continuous blood inflow is maintained with no loss of kinetic energy, thus increasing the circulation. The pneumatic circuit manufactured that drives the pulsatile pump via pneumatic pressure consists of air inlet and outlet solenoid valves. The air pump delivers the blood in the sac contracted by oxygen pressure to the oxygenator through the outlet solenoid valve. To drive one pulsatile pump, six solenoid valves were connected according to the control algorithm. Oxygen from the outer case must be ejected quickly because a blood sac that has contracted after ejecting blood must be inflated before contracting again, to prevent damage to the blood sac. An additional air pump can be installed between the solenoid valve and oxygenator to increase the blood suction pressure through negative pressure. The ECMO device control unit has 12 solenoid valves, an internal air pump, and an LCD touch screen that controls the operation of the device. The pulsatile pump operated by the control unit can be adjusted from 80 to 120 bpm.

2.2 In-vitro devices and settings

The evaluation system used for the in-vitro experiment consisted of a clamp, compliance chamber, and reserve container, and was connected to the pulsatile ECMO device. The compliance chamber controlled the compliance and generated a pressure change from 80 to 120 mmHg, mirroring arterial pressure, to achieve pulsating blood flow with a single 50-100 cc stroke. The fluid flowing into the compliance chamber could be released through the clamp, and the chamber was sealed to prevent leakage. The reservoir container was designed to store the fluid flowing through the clamp and maintain a pressure of 10-20 mmHg, which is the same as the left atrial pressure in the human body. The pressure in the reservoir was determined by the difference between the inflow from the compliance chamber and the outflow into the ECMO device. If the circulating flow in the ECMO device is not kept constant, the reservoir pressure increases abnormally. The clamp between the compliance chamber and the reservoir was designed to control fluid resistance by partially closing or opening the fluid path, and to adjust the average pressure in the compliance chamber to 100 mmHg, which equals mean arterial pressure. As shown in Fig. 2, the pressure in the compliance chamber was measured directly by a sensor (MPXV2202DP; Freescale Semiconductor, USA) and recorded on a computer using LabVIEW and a USB-6001 data-acquisition (DAQ) device (USB-6001, National Instruments, USA). The outflow rate of the ECMO device was measured using a probe at the ECMO device inlet and an ultrasonic blood flow meter (TS410, Transonic, USA) at the outlet, and recorded simultaneously using the DAQ device.

2.3 Experimental conditions

Preload is influenced by the patient’s central venous pressure (CVP) and the relative height difference between the patient and ECMO device. The reservoir pressure, which reproduces the CVP, can be adjusted by adding more fluid to, or draining fluid from, the MOCK system reservoir. To increase the inlet pressure and blood inflow of the pulsatile ECMO device, the ECMO device pump should be located below the patient. However, it is necessary to examine how the ECMO device inlet pressure affects the ECMO device output, since the heights of the ECMO device and the patient can differ when transferring the patient, such that the patient’s CVP can decrease. When the average ECMO device inlet pressure was maintained at 40 mmHg, the pump rate was set to 80, 100, and 120 bpm, and the average pressure in the compliance chamber was adjusted to 80, 100, 120, and 140 mmHg. The ECMO device outflow was measured at each setting. Then, the ECMO device preload was lowered to 10-30 mmHg under the same conditions, and the ECMO device outflow was recorded. To overcome the reduced ECMO device outflow at low preloads, another small air pump was installed between the membrane oxygenator and the solenoid valve initially connected to the membrane oxygenator. Then, the ECMO device outflow was measured and compared to the previous results.

3. Results

3.1 Output of the pulsatile pump at a high constant oxygen pressure in the ECMO device

Fig. 3 shows the pulse waveforms of blood pressure (afterload of the ECMO device) and blood flow measured by the MOCK system in vitro experiment. As an index of pulsatility, EEPI was calculated according to the power obtained from changing the pressure and flow during a pulse period. The power was obtained from the mean pressure and flow [18]. While the existing pulsatile centrifugal ECMO has low pulsatility with an EEPI of 5 %, the suggested ECMO has sufficient pulsatility, with an EEPI of 23.3-24.4 %.

Fig. 4 shows the measured ECMO device outflow under various conditions when the oxygen tank pressure was used to power the ECMO device. The oxygen pressure in the oxygen tank was increased to 120 bar, and the pressure was adjusted in stages using two regulators connected to the oxygen tank. The first regulator (CPR-OGR 550, CPR Korea, Korea) limited the O2 outflow to below 10 L/min, and the other lowered the O2 pressure to 2.1 bar, which was suitable for the power. The medical oxygen tank (Medical Cylinders, Maltanimetals, Korea) used in this study contained 4.6 L of 98 % oxygen.

The stroke volume of the ECMO device decreased as the afterload increased, and the blood flow increased with the input pressure. As shown in Fig. 3, the ECMO device outflow could be adjusted using the pump rate; when the pump rate was increased to 120 bpm, the blood flow increased to 5 L/min.
However, when the inlet pressure was lowered, the blood flow decreased to 3.5 L/min. Various factors can reduce the inlet pressure, including improper positioning of the ECMO device. This must be corrected immediately to increase inflow to the ECMO device, and the patient's blood volume should maintain the inlet pressure without adjustment, to avoid damaging blood vessels. The 120-bar oxygen tank was able to drive the ECMO device for 50 min when the ECMO device outflow was maintained at 5 L/min.

3.2 Role of the air pump

As shown in Fig. 5, an air pump was connected to the solenoid valves to discharge oxygen from the pulsatile ECMO device, to drain oxygen from the blood sac without having to adjust the input pressure due to the patient's position. Fig. 5 shows that when the pump operates at 100 or 120 bpm, the flow rate was insufficient due to the decreased flow under the same ECMO device ejection conditions in the MOCK circulation system. When the air pump operated, the ejection rate increased because the pump generated negative pressure on the blood sac to induce sufficient blood flow. This had little effect on the pump flow rate, but greatly increased the output. The air pump attached to the ECMO device had an output of 5 watts, measured 60×120 mm, and weighed less than 200 g.

4. Discussion and conclusions

This study used the EEPI, which is the ratio of the increased hemodynamic energy of pulsatile flow to the total hemodynamic energy of continuous flow, at the same average flow rate and pressure, as a criterion for evaluating the pulsatility of the
The ECMO proposed here could maintain sufficient blood flow of 5 L/min, and blood pressure of 140 mmHg, and also had sufficient pulsatility with an EEPI of 20 % in-vitro experiment using the MOCK system. However, the MOCK system cannot simulate the patient's perfusion resistance, which decreases under pulsatile flow but can be observed in-vivo experiment. If a continuous flow pump is applied to the MOCK system under the same resistance, to compare the hemodynamic characteristics when using the suggested pulsatile ECMO, the mean blood flow and mean blood pressure of the pulsatile pump should be the same due to the constant perfusion resistance of the MOCK system. However, if the resistance of the MOCK system is arbitrarily kept low, the maximum blood flow and pressure can exceed that of the pulsatile pump, because a continuous flow pump is more efficient than a pulsatile pump.

Since pulsatile blood flow decreases the patient's perfusion resistance, previous clinical trials showed that the flow output of continuous pumps decreased. Therefore, the hemodynamic characteristics of the proposed pulsatile ECMO, hemolysis, and degree of thrombus formation must be evaluated in animal experiments. This will facilitate research on the effects of pulsatile blood flow by enabling clinical trials, since the suggested ECMO overcomes the limitations of existing pulsatile pumps.

**Acknowledgments**

This research was supported by the Basic Science Research Program through the National Research Foundation of Korea (NRF) funded by the Ministry of Education (2020R1A4A1019475). This work was supported by the Korea Medical Device Development Fund grant funded by the Korea government (the Ministry of Science and ICT, the Ministry of Trade, Industry and Energy, the Ministry of Health & Welfare, the Ministry of Food and Drug Safety) (Project Number: 202013A02-02).

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