Venous cannula performance assessment in a realistic caval tree model

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Abstract

OBJECTIVES: A new caval tree system was designed for realistic in vitro simulation. The objective of our study was to assess cannula performance for virtually wall-less versus standard percutaneous thin-walled venous cannulas in a setting of venous collapse in case of negative pressure.

METHODS: For a collapsible caval model, a very flexible plastic material was selected, and a model with nine afferent veins was designed according to the anatomy of the vena cava. A flow bench was built including a lower reservoir holding the caval tree, built by taking into account the main afferent vessels and their flow provided by a reservoir 6 cm above. A cannula was inserted in this caval tree and connected to a centrifugal pump that, in turn, was connected to a reservoir positioned 83 cm above the second lower reservoir (after-load = 60 mmHg). Using the same pre-load, the simulated venous drainage for cardiopulmonary bypass was realized using a 24 F wall-less cannula (Smartcanula) and 25 F percutaneous cannula (Biomedicus), and stepwise increased augmentation (1500 RPM, 2000 and 2500 RPM) of venous drainage.

RESULTS: For the thin wall and the wall-less cannulas, 36 pairs of flow and pressure measurements were realized for three different RPM values. The mean Q-values at 1500, 2000 and 2500 RPM were: 3.98 ± 0.01, 6.27 ± 0.02 and 9.81 ± 0.02 l/min for the wall-less cannula (P <0.0001), versus 2.74 ± 0.02, 3.06 ± 0.05, 6.78 ± 0.02 l/min for the thin-wall cannula (P <0.0001). The corresponding inlet pressure values were: −8.88 ± 0.01, −23.69 ± 0.81 and −70.22 ± 0.18 mmHg for the wall-less cannula (P <0.0001), versus −36.69 ± 1.88, −80.85 ± 1.71 and −101.83 ± 0.45 mmHg for the thin-wall cannula (P <0.0001). The thin-wall cannula showed mean Q-values 37% less and mean P values 26% more when compared with the wall-less cannula (P <0.0001).

CONCLUSIONS: Our in vitro water test was able to mimic a negative pressure situation, where the wall-less cannula design performs better compared with the traditional thin-wall cannula.

Keywords: Caval collapse • Atrial chatter • Venous drainage • Cardiopulmonary bypass • Venous cannula

INTRODUCTION

Cardiopulmonary bypass (CPB) is commonly used during heart surgery to maintain the circulation of blood in the body during heart arrest. Likewise, the circulation of blood in extracorporeal life support is maintained with a pump oxygenator [1, 2]. In both situations, drainage of the systemic venous blood remains a concern [3–5]. Inadequate venous drainage during open heart surgery can disturb the physiological flow of blood, giving rise to intermittent atrial and/or caval collapse and, consequently, so-called ‘atrial chatter’ or even complete shut-off of venous drainage. Although the phenomenon described has always been present and observed since the invention of CPB, the efforts made for prevention are scarce and better understanding is essential for the development of improved perfusion strategies.

During CPB, the amount of systemic venous drainage overall is directly correlated with the amount of pump flow. As a consequence, the pump flow is limited by the amount of venous blood that the pump is receiving [6]. Several factors can influence the quality of venous drainage during CPB, including pre-load, after-load, pump design [7–11] and venous cannula design. In addition, the flexibility of the venous system induces a further parameter due to the veins can collapse under negative pressure (suction) partially or totally. Hence, even in the presence of sufficient circulatory blood volume, blood may not reach the pump because of venous collapse.

The design and the positioning of the venous cannula can influence the quality of venous drainage during CPB. Improved blood drainage can be achieved, as previously demonstrated for the virtually wall-less cannula (www.smartcanula.com; Smartcanula LLC, Lausanne, Switzerland), which is based on the ‘collapsed insertion and expansion in situ’ principle [4, 12–14].

This cannula is essentially a circumferential wire structure in which the entire intravascular part allows for drainage (Fig. 1). In addition, this device with its self-expanding design acts also as a spacer preventing the vein from collapsing and, therefore, allows all collateral blood to be drained directly towards the pump oxygenator. This contrasts with the current design of the commercially available thin-wall cannulas, which have discrete holes in the plastic tubing allowing for drainage at specific spots.
Originally, the wall-less cannula design (e.g. 36 F) was developed for venous drainage by gravity. However, in minimally invasive surgery, augmentation with constrained force centrifugal pumps or vacuum is often used. Likewise, augmentation is standard in extracorporeal life support (ECLS) and extracorporeal membrane oxygenation (ECMO) for cardiac and/or pulmonary assist. Therefore, smaller (e.g. 24 F) wall-less cannulas can be used in these settings.

We recently designed a simplified caval tree bench model [15] to mimic caval collapse originating from excessive venous drainage in a reproducible fashion on the bench for better understanding of how it is generated and hopefully prevented. In the present study, we examined the impact of the collapsible vena caval tree model on the flow dynamics for traditional thin-wall cannulas versus virtually wall-less cannulas designed for use with augmentation.

MATERIALS AND METHODS

Manufacture process of the caval tree model

The geometry of the venous tree for the caval model was compiled as previously described in detail [15]. Briefly, a simplified caval tree representing nine major afferent veins and one large efferent channel is cut by a computer-controlled laser system (Versa laser, Universal Laser Systems, Scottsdale, AZ, USA) simultaneously from two stacked polyethylene layers. The two layers of the plastic material are simultaneously sealed tight by automatic welding during the cutting process. This allows for a seamless and flexible connection of the afferent channels to the caval axis similar to that in nature.

Experimental set-up

The experimental set-up was designed as previously described in detail [15]. Briefly, the afferent veins of the caval model are connected with feeding reservoirs (F1 and F2) as shown in Fig. 2, which provide in an overflow mode a constant inflow at a defined height of 6 cm, resulting in a preload of 6 cmH2O as a function of their height relative to the caval axis. A thin-wall venous cannula (3/8” 25 F Biomedicus, Medtronic, Minneapolis, USA: control) versus a virtually wall-less cannula (3/8” 24 F 530 mm ST Smartcanna, Lausanne, Switzerland) is inserted axially into the caval tree and snared at the point of insertion. The caval tree model with the cannula inside is positioned underwater (2 cmH2O) in the immersion reservoir (I). The latter is required to provide a water seal to prevent aspiration of air from the environment. Air would interfere negatively with the performance of the centrifugal pump. The overflow from reservoirs F1 and F2, and reservoir (I) drops into collecting reservoir (C), (65 cm long, 47 cm wide and 30 cm high; KAISER + KRAFT/St-Sulpice, Switzerland) and is continuously pumped to an arterial reservoir (A) of the same size to maintain the defined immersion of the caval tree. Reservoir (A) is positioned at 83 cm above the caval axis by means of a memory-programmed RK Easy lift elevator (Rose + Krieger, Minden, Switzerland), thus providing a physiological after-load of 60 mmHg for the cannula-centrifugal pump set-up connected to the reservoir (A).

Sensors

Pressure at the cannula outlet was measured at the centrifugal pump inlet (Fig. 2) using Millar pressure transducers (Millar Instruments, Inc., Houston, TX, USA) sampled at 1 KHz and connected to a data acquisition board (National Instruments Corporation, Ennetbaden, Switzerland). Baseline conditions for this sensor were established by calibrating the pressure value to be equal to zero in the absence of flow in the test tube. Additionally, a clamp-on ultrasound flow probe was attached to the test tube (Transonic Systems, Inc., Ithaca, New York, NY, USA) to measure the flow rate in this tube. The ultrasound flow meter, calibrated using a volumetric tank and a timer, was also connected to the acquisition board and sampled at 1 kHz.

Data acquisition application

Pressure and flow sensors were controlled through a custom-made application using the LabView 8.2.1 software (National Instruments, TX, USA). The cannulas were tested at three reproducible pump speeds without collapse (1500, 2000 and 2500...
RPM) to obtain complete data sets for both cannula types at all measuring points. Computerized sampling was realized for each pump speed condition, and the application measured the pressure and flow rate during 20 s. Measurements were then averaged over the capture period and the average data were stored to be later analysed using Microsoft Excel and Graph Pad Prism.

Statistical analyses

The unpaired Student t-test was used to compare negative pressure and flow rate between the wall-less and standard cannulas for the three pump speeds. The two-way analysis of variance (ANOVA) test was also used to determine whether the type of drainage at the three pump speeds (1500, 2000 and 2500 RPM) was a significant factor affecting the negative pressure and flow rate measurements. These ANOVA tests were done separately for the wall-less and control cannulas. For all statistical tests, the significance level was set a priori at $P < 0.05$.

RESULTS

For the thin-wall and the wall-less cannulas, 36 pairs of flow (Q) and pressure (P) measurements were realized for three different RPM. The mean Q-values at 1500, 2000 and 2500 RPM (Fig. 3) were: $3.98 \pm 0.01$, $6.27 \pm 0.02$ and $9.81 \pm 0.02$ l/min for the wall-less cannula versus $2.74 \pm 0.02$, $3.06 \pm 0.05$ and $6.78 \pm 0.02$ l/min for the thin-wall cannula ($P < 0.0001$). The corresponding inlet pressure values (Fig. 4) were: $-8.88 \pm 0.01$, $-23.69 \pm 0.81$ and $-70.22 \pm 0.18$ mmHg for the wall-less cannula, versus
\[ -36.69 \pm 1.88, -80.85 \pm 1.71 \text{ and } -101.83 \pm 0.45 \text{ mmHg} \text{ for the thin-wall cannula (}\bar{P} < 0.0001). \]

Fig. 5 shows the relation between flow rate and negative pressure in the wall-less and the thin-wall cannula. This figure shows that there is an increase in the flow rate with negative pressure at the different RPM for both cannulas. The thin-wall cannula showed mean \( Q \)-values 37% less and mean \( P \)-values 26% more when compared with the wall-less cannula (\( \bar{P} < 0.0001 \)). Figure 6 shows the interesting linear relation between the log flow rate and the inlet negative pressure.

**DISCUSSION**

Using our realistic caval tree model, wall-less cannulas have a better flow rate for each RPM with a mean increase of 37% compared with thin-wall cannulas (Fig. 3). Superior venous drainage with the wall-less cannula has previously been studied in vitro and in vivo with gravity drainage without a centrifugal pump [16, 17]. However, these results were based mainly on the superior diameter of the self-expanding design. Several reasons explain the better blood drainage with a wall-less cannula with increased hoop strength designed for use with augmentation in the current setting. First, the virtually wall-less cannula allows for direct blood drainage at all levels of afferent vessels (e.g. iliac, renal, hepatic etc.). Second, the cross-sectional area within the vessel lumen is still larger for the wall-less design compared with a percutaneous thin-wall cannula. Finally, the virtually wall-less cannula has effectively a thinner wall (between 0.0 and 0.36 mm) compared with the thin-wall percutaneous cannula (>0.60 mm), a difference that does affect the effective cross-sectional area.

Figure 4 describes the inlet negative pressure of the cannulas. It shows the superiority of the wall-less cannula with less negative pressure compared with the thin-wall cannula for each RPM (\( \bar{P} < 0.0001 \)). The wall-less cannula has tremendous potential for use. High negative pressure in the thin-wall cannula increases haemolysis, and the risk of air aspiration, cavitation and embolism formation. The target blood flow for CPB is 2.4 l/min/m\(^2\). For an average patient, this accounts typically for \( \sim 4 \text{ l/min} \), a target that is reached easily with a wall-less cannula requiring a negative pressure around –5 mmHg. Our results demonstrate the superiority of the wall-less cannula design when a centrifugal pump is added, which might permit a reduction in the cannula diameter and achievement of the theoretical blood flow without difficulty.
A centrifugal pump has been used for a long time [18, 19] in peripheral venous drainage. In 1999, Taveaerai's study compared different venous cannulas in vivo with a central venous pressure of 10 mmHg and showed that to reach a 4 l/min blood flow, negative pressures of −78 mmHg for a 24 F cannula and −59 mmHg for a 28 F cannula were needed [20]. A more recent study measured venous drainage for cardiopulmonary support and for ECMO. It showed that the theoretical blood flow was not reached with 20–24 F thin-wall cannulas and highlighted the risk of chattering of the vein, which can lead to obstruction of the cannula holes [20]. A cardiopulmonary support system was tested in vivo with a 21 F thin-wall cannula and the result was that a negative pressure of −102 mmHg was needed to reach a blood flow of 3.9 l/min, which is not sufficient for a full flow [21]. This finding is very much in line with the observation in our study where the 24 F thin-wall cannula required −90 mmHg for a flow ∼4.0 l/min.

In our study, we used a wall-less cannula with active drainage by a centrifugal pump. Our experiment was carried out in vitro and we demonstrated the superiority of the wall-less cannula compared with a control thin-wall cannula. The theoretical blood flow for a CPB in cardiac surgery can be reached with a negative inlet pressure of −5 mmHg with a 24 F wall-less cannula. Our results are further supported in vivo by two studies. The first one showed that, with an 18 F wall-less cannula, the theoretical blood flow in animals going through CPB was achieved with a negative pressure less than −50 mmHg [22]. The second study was also made in vivo using a portable extracorporeal oxygenator and a wall-less cannula. The results showed that, with a negative pressure of −43 mmHg, the theoretical blood flow could be reached with an 18 F access and a wall-less cannula [23]. In another in vivo study [24], an increase in blood flow with a 25 F access and a wall-less cannula was 43% compared with the one reached with a 25 F thin-wall cannula.

In our study, the increase in flow rate was slightly smaller, 37%, which is still significant, but using a smaller wall-less cannula. As a limitation of this study, it can be argued that water was used to prime the circuit. However, for such a large cannula diameter, no relevant difference is observed at clinical flow rates (personal communication LK von Segesser). Furthermore, this systematic simplification is the same for the test cannula and the control.

With regard to the caval collapse mode observed, it is important to distinguish between the two modes described earlier. For the 25 F thin-wall cannula, the aspiration of the caval wall substitute into the orifices at the cannula tip occurs at negative pressure >120 mmHg. We have previously studied this question for the 23 F thin-wall cannula where quite sudden complete shut-off of drainage occurred at −110 mmHg. Usually, complete pump-stop is required to release the grip of the venous wall, before the flow can progressively be recovered up to a threshold negative pressure below the collapse level. For the 24 F virtually wall-less cannula design, the flow is close to 10 l/min at −70 mmHg and, therefore, caval collapse cannot be studied in this set-up. However, we do know from previous studies with this 3/8" 24 F 530 mm ST wall-less cannula configuration that, at excessive negative pressure below −120 mmHg, the caval wall substitute will be attracted progressively all over the cannula length and compression of the self-expanding structure with increasing reduction of venous drainage occurs also for the virtually wall-less cannula design. However, somewhat different is the release mechanism, because reduction of negative pressure results in progressive re-expansion of the self-expanding design and, thus, recovery of venous drainage.

Our in vitro test was able to mimic a negative pressure situation, where the wall-less cannula design performs better compared with a traditional thin-wall cannula. With the virtually wall-less cannula design, smaller cannula diameters are possible and reduced negative pressures are needed for assisted blood water drainage. The application of these findings could contribute to further improvement of perfusion in the clinical setting.

Conflict interest: none declared.

REFERENCES


