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Modeling Systolic Pressure Variation Due to Positive Pressure Ventilation

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Abstract— **Although many clinical techniques have been proposed to assess blood volume none have been established as an undisputed standard practice [2]. Volume studies suggest systolic pressure variation (SPV) as a promising volume indicator but underlying influences on SPV are not well understood. Successful modeling of SPV will reveal the major SPV influencers, guide algorithm development to accommodate these influencers, and potentially lead to a more clinically relevant interpretation of SPV values, thus improving upon current clinical methods for assessing blood volume. This study takes a first step towards identifying SPV influencers by investigating three variations of an existing pressure-flow cardiovascular model [9]. Each successive version introduces an additional modification in attempt to model SPV under normovolemic and hypovolemic conditions, where the last model accounts for positive pressure ventilation, venous compression, and a rightward septum shift. Under normovolemic conditions, each model yields SPV values of 5.8, 6.4, and 6.7 mmHg, respectively. Under hypovolemic conditions the results do not agree with clinical findings, suggesting these three mechanisms alone do not dictate the clinical SPV response to a decrease in volume. Model results are used to suggest improvements for future work.**

I. Introduction

TODAY patient fluid status is best assessed by considering multiple sources and integrating each to make a clinical decision. For example, doctors may consult transesophageal echo, stroke volume, blood flow, capillary refill time, and/or pressure variations to determine whether or not to administer fluid. Any one of these measurements alone is insufficient since each is influenced by a number of factors besides blood volume [2,7]. If one of these measurements is to emerge as a gold standard, we must understand its relationship to its influencing factors and adjust the clinical interpretation accordingly.

Understanding the underlying mechanisms for systolic pressure variation (SPV) during positive pressure ventilation (PV) has been the focus of many previous studies. Most of these agree that an enhanced SPV during PV results from changes in pressure gradients between the intrathoracic and extrathoracic cavities. An increase in intravascular pressure in the chest cavity decreases left ventricular afterload by increasing the pressure gradient between the aorta and systemic vasculature and decreases venous return by decreasing the pressure gradient between the vena cavae and the right atrium. The overall effect is an initial increase in arterial pressure (delta up) followed by a decrease in arterial pressure (delta down) for each respiration cycle [6]. In addition, studies have shown compression of vasculature, such as the vena cavae, during PV may contribute to a reduced venous return [10], while a rightward shift of the septum may increase ventricular preload, thus increasing cardiac output [4].

The goal of this study was to capture each of these underlying mechanisms by modifying an existing pressure-flow model of the cardiovascular system in attempt to accurately model the effects of PV on SPV under normovolemic as well as hypovolemic conditions.

II. Method

The established model chosen for this study is a single loop resistor-inductor-capacitor (RLC) circuit with multiple RLC segments, each representing a major vasculature group within the cardiovascular (CV) system (fig. 1). This model was chosen to balance simplicity with fidelity. The single loop model yields beat-to-beat waveforms with moderate fidelity and represents the major systems mentioned most often in SPV literature. On the other hand, its simplicity allows for quick simulation time and linear relationships.

Circuit component values were derived based on fluid mechanics; assuming vessels are cylindrical with linear elasticity [9]. All components are static except for the left ventricular compliance (or stiffness, where compliance = 1/stiffness), which is modeled as a rectified sine wave. This periodic left ventricular stiffness is meant to reflect systolic and diastolic changes and, most importantly drives pulsatile outputs. The periodicity of this stiffness function is 0.8 sec, resulting in a simulated heart rate of 75 beats/minute.

The first modification made to this model includes replacing the capacitor grounds in the chest cavity with an airway pressure waveform and allowing 20% of the airway pressure to transmit to the vascular system. Actual pressure transmission and its distribution within the chest cavity is not well known and will vary depending on individual patient compliance [10]. The simulated airway pressure assumes a respiration rate of 10 breaths/min, an I:E ratio of 1:1, and is meant to resemble measured airway waveforms during volume-targeted controlled ventilation [3]. The function used to model this ventilation is the positive portion of a periodic sine wave skewed by the addition of its second harmonic (fig. 2).

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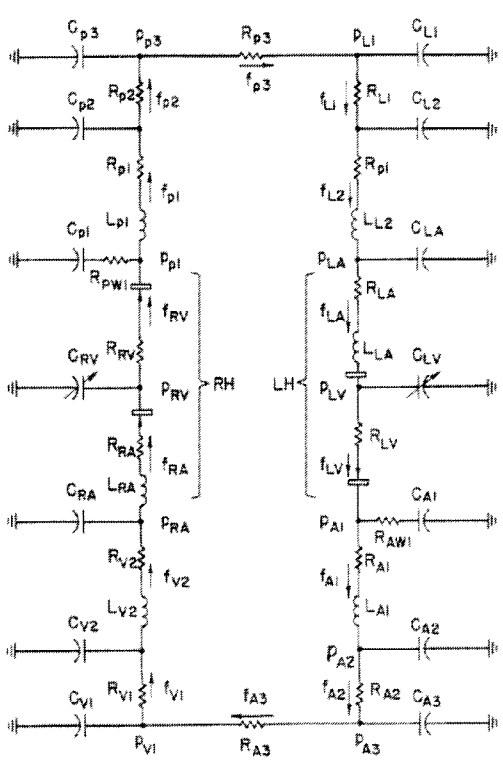

Fig. 1. Rideout's single loop pressure-flow cardiovascular model. Starting from the segment in the center on the left side, flow proceeds from the right heart, to the pulmonary arteries, pulmonary veins, the left heart, aortic artery, systemic arteries, the systemic veins, and finally back to the right heart.

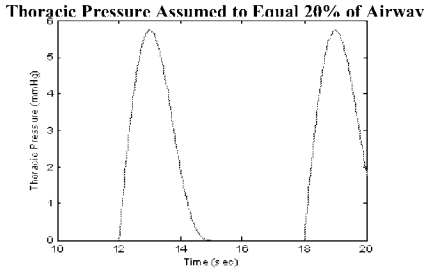

Fig. 2. Rectified sine wave skewed by the addition of its second harmonic. Applied to the ground terminal of the capacitors in the chest cavity.

The second version of the model attempts to model vena cavae compression during PV by introducing a variable resistance in the systemic venous compartment, which increases with intrathoracic pressure.

Lastly, the third version considers the proposed rightward septum shift during PV. This phenomonen was measured by Mitchell via ultrasonic transducers and showed that PV reduces right ventricular end diastolic volume (RVEDV) causing a rightward shift of the septum [4]. This shift enhances left ventricle end diastolic volume and leads to an increase in stroke volume. An increase in stroke volume results in a larger delta up and ultimately an increase in SPV. To model this effect, version 3 of the model compares the RVEDV value to a reference value taken at normovolemia and apnea. If the current RVEDV is less than the reference, the left ventricular stiffness function is multiplied by a scaling factor (equation 1), thus simulating an increase in cardiac pumping strength and, subsequently, stroke volume.

$$LVS = 1 + \frac{(RVEDVref - RVEDV)}{RVEDVref} \qquad \text{(equation 1),}$$

where RVEDVref is 150ml, and LVS is the left ventricular stiffness scaling factor. LVS is bound between 0.1 and 1.9, bounding stroke volume between 60ml and 67ml.

All three models were run for 60 seconds at a frequency of 100Hz under normovolemic conditions (5 liters of blood). Once in steady-state (after ~10sec) SPV measurements were calculated by subtracting the minimum systolic pressure from the maximum systolic pressure in each respiratory cycle. Delta up was calculated as the maximum systolic pressure minus systolic pressure during apnea in one respiratory cycle. Delta down was calculated as systolic pressure during apnea minus minimum systolic pressure during a respiratory. Measurements were calculated for each breath cycle and the median over 5 cycles was used for comparison to clinical results.

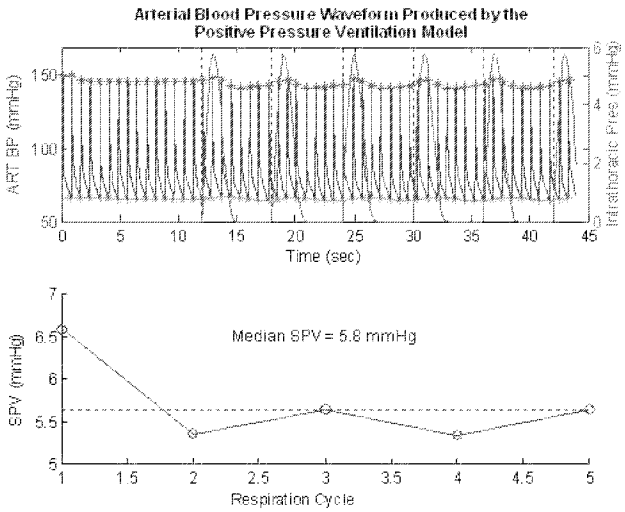

Fig. 3. (Top) Arterial waveform produced by the model including PV. Asterisks at the peak and trough of the waveform mark systolic and diastolic values, respectively. Respiration cycles are separated by vertical dotted lines. (Bottom) SPV values corresponding to each breath cycle

Hypovolemic conditions were simulated by incrementally removing 100ml of fluid until 800ml of fluid loss was achieved. To investigate the model's sensitivity to individual circuit components, the model including PV only was run multiple times with incremental changes in value for one component while all other values remained constant.

III. Results

The cardiovascular model with 20% airway pressure transmitted to the intrathoracic cavity yields SPV values similar to that described in literature under normovolemic conditions [5,6,8] (Table 1). For this model the SPV

calculated at the onset of the steady-state (t = 12 sec), is 5.8 mmHg, delta up is 1.7 mmHg, and delta down is 4.1 mmHg (Fig 3). Experimental results categorize patients with an SPV of 9.5 +/-4.6 mmHg and delta down of 4.9 +/-3.3 as non-fluid responders, or normovolemic [5,8].

In the model accounting for vena cavae compression, the vascular resistance increases with positive pressure causing the minimum systolic pressure during a breath to drop when compared to the model with static resistance. During normovolemia, the result is an increase in delta down which increases SPV from 5.8 to 6.4 mmHg.

Lastly, during normovolemia, SPV increases from 6.4 to 6.7 mmHg when left ventricular stiffness is allowed to change with right ventricular volume.

Under hypovolemic conditions, the PV only model and the PV plus variable vena cavae resistance model show a slight decline in SPV as fluid loss approaches 800ml at approximate rates of -0.4mmHg/L, and –0.8mmHg/L, respectively. The model including PV with dynamic vena cavae resistance and dynamic ventricular compliance yields an approximately flat SPV curve up to a blood loss of 300ml and then shows a slight rise in SPV, approximately +1.2mmHg/L, as blood loss continues to 800ml.

	Normovolemic		Hypovolemic (~500ml loss)	
(mmHg)	SPV	Δ down	SPV	Δ down
PV only	5.8	4.1	5.6	3.8
DVR	6.4	4.6	6.0	4.1
DVS	6.7	4.5	7.0	4.5
Clinical	9.5 +/-4.6	4.9 +/-3.3	14.3 +/- 6.5	8.4 +/-4.8

Table 1. Model results for SPV during normovolemia and hypovolemia compared to clinical data in literature [5,6,8]. VVR refers to the model with dynamic vena cavae resistance and VVS refers to the model with dynamic ventricular stiffness. (Δ = Delta)

IV. Discussion

The models' accurate physiological representation of SPV under normovolemic conditions is understood by considering the changes in pressure gradients due to PV. As intrathoracic pressure increases, blood pressure in thoracic vessels increase equally, assuming equal pressure distribution throughout the cavity. Thus, pressure gradients between vasculature within the thoracic cavity remain unchanged. On the other hand, the pressure gradients between the intra and extra thoracic vessels will increase on the arterial side and decrease on the venous side, in the direction of blood flow. These pressure gradients result in an initial increase in cardiac output (delta up) and an eventual pooling of blood in the veins resulting in a subsequent decrease in cardiac output (delta down). Positive thoracic pressure in the PV only model increases venous volume from 3.67 to 3.86 liters. Due to high venous compliance this increase in volume does not significantly increase venous pressure. Thus, the pressure gradient between the venous intra and extra thoracic

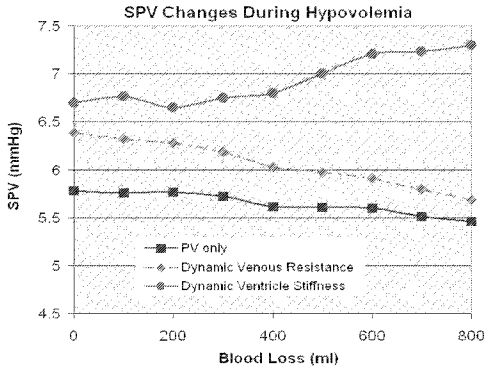

Fig. 4. SPV slightly decreases with blood loss in the PV only and dynamic venous resistance models. SPV shows a slight increase in the dynamic ventricular stiffness model, agreeing with trends found clinically.

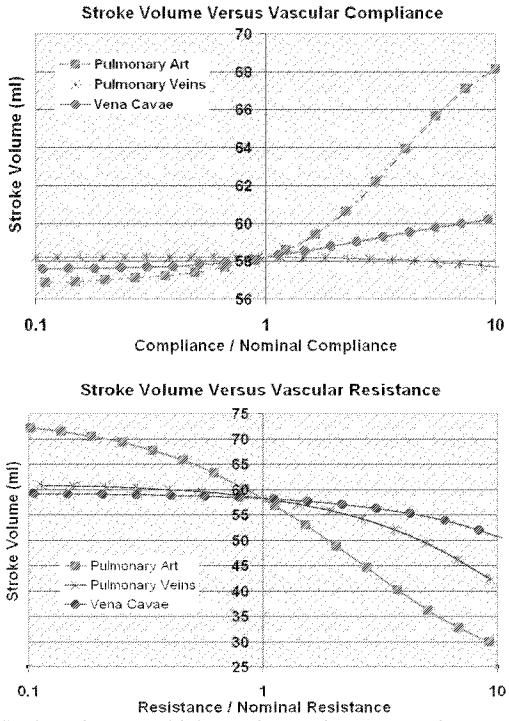

Fig. 5. Stroke volume sensitivity to changes in venous resistance (top) and compliance (bottom).

segments is reestablished only when the thoracic pressure is released. Of course this same analysis translates into terms of voltage and current, where all relationships described in terms of pressure and flow remain valid. Thus, the RLC model is able to elicit an appropriate SPV response under normovolemic conditions.

Under PV, the pulmonary vasculature and the systemic veins, or the vena cavae, are most likely to undergo radius changes due to vessel compression [10]. Since vessel resistance is proportional to the fourth power of vessel radius [9], slight changes in radius have a significant impact on resistance. As demonstrated by the model with dynamic resistance, a 60% change in vena cavae resistance (10 to 16 gm/sec*cm4) caused a 10% change in SPV (5.8 to 6.4 mmHg) during normovolemia. Although the dynamic resistance model did not show an increase in SPV with fluid

loss this model suggests the importance of accounting for any radius changes during fluid loss and PV. Such changes may become exaggerated under hypovolemic conditions [1], resulting in a higher SPV, which would agree with the expected clinical trend. Thus, a future improvement may be to make resistance variation a function of blood volume as well as thoracic pressure.

As shown in figure 4, the PV only model and the PV with dynamic resistance model do not demonstrate an increase in SPV due to volume loss. Instead, as blood volume drops by 800ml SPV falls from 5.8 to 5.5 mmHg and 6.4 to 5.7mmHg, respectively. Despite a drop in blood pressure with the loss of blood, the pressure gradient between intra and extra thoracic vasculature still equals the applied thoracic pressure. Since the pressure gradients are the same in both a normovolemia and hypovolemia case, SPV is also about the same. The slight decline in SPV is attributed to a small decrease in stroke volume variation (peak value minus the minimum) due to less energy stored by vessel elasticity as blood volume drops throughout the elastic compartments.

In the third model, varying left ventricular stiffness according to equation 1 results in an increase in SPV, compared to the first two models, due to an increase in stroke volume variation. Here stroke volume periodically varies from 60ml to 67ml compared to 61ml to 65.5ml for the first two models. As blood loss approaches 300ml, RVEDV is minimally affected, thus SPV is fairly constant. After 300ml of blood loss, RVEDV starts to decline, causing stroke volume variation to increase, and subsequently SPV. Thus, this last version of the model demonstrates a slight increase in SPV as blood volume drops. However, the overall increase in SPV by 0.5 mmHg is considerably less than the increase of 4.8 mmHg quoted in the literature for a drop of 500 ml of blood volume [8]. This suggests there are additional mechanisms contributing to an increase in SPV during hypovolemia.

In attempt to understand other possible contributing factors, model sensitivity to resistance and compliance changes were explicitly evaluated. Results suggest slight changes in pulmonary compliance and resistance during PV or blood loss will significantly impact calculated stroke volume (fig. 5). Thus, future models may introduce dynamic pulmonary components to more accurately capture SPV.

V. Conclusion

The first version of the SPV model attempts to mimic clinical SPV results by solely considering pressure-flow relationships in the presence of PV. As pressure gradients change with thoracic pressure the model produces delta up, delta down and SPV values that agree with clinical values reported in literature for normovolemia. However, it does not accurately depict changes due to loss of blood volume. Thus, this pressure-flow model with PV must not model all factors SPV depends on.

Since resistance changes are likely to occur during mechanical ventilation due to vessel compression [5], version 2 of the SPV model incorporates a variable venous resistance. The result of the variable resistance is an increase in SPV for a normovolemic condition. But, again version 2 does not display an increase in SPV as the system loses fluid. However, results do support including variable resistance which is a function of applied pressure as well as blood volume in future models.

In the third version of the SPV model left ventricular diastolic stiffness is a function of RVEDV to represent the interaction between the left and right heart. As volume drops by 800 ml, SPV increases by 0.5 mmHg. This amount accounts for only about 10% of the clinically determined increase of 4.8mmHg, suggesting additional mechanisms influencing SPV are at play.

Improved models may incorporate changes in vascular tone associated with fluid loss by modeling the baroreceptor response [9]. Next, vessel compression, or resistance, should be modeled as a function of transmural pressure. Later, more complex system changes such as redistribution of blood, i.e. shunting of blood away from the periphery, and re-absorption of fluid from tissue could be introduced.

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