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A model of perfusion of the healthy human lung

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ABSTRACT

This study presents a model that simulates the pulmonary capillary perfusion. The model describes the lungs as divided into horizontal layers and includes: capillary geometry; capillary wall elasticity; pressure at the pulmonary artery; blood viscosity; the effect of the chest wall; the change in lung height and hydrostatic effects of the lung tissue and of the blood during breathing. The model simulates pulsatile blood perfusion with an increasing blood distribution down the lungs, in agreement with previous experimental studies. Moreover the model is in agreement with experimentally measured total capillary perfusion, total capillary volume, total capillary surface area and transition time of red blood cells passing through the pulmonary capillary network. The presented model is the first to be validated against the mentioned experimental data and to model the link between airway pressure, lung volume and perfusion.

1. Introduction

Appropriate ventilator settings for intensive care patients with respiratory disorders are crucial for reducing recovery time and minimizing the risk of ventilator induced lung injury (VILI) [1,2]. Finding the appropriate settings requires a trade-off between the need to obtain adequate gas exchange and minimization of VILI. Positive end expiratory pressure (PEEP) and tidal volume affect both the lung mechanics and the gas exchange, and appropriate levels for example of tidal volume have been shown to lower the mortality in patients with acute lung injury [3]. Furthermore high pressures by means of PEEP have proven to prevent alveolar collapse and improve gas exchange in the diseased lungs [2]. However, it is still not clear how to improve gas exchange while preventing VILI [2]. In order to understand gas exchange in patients with respiratory failure, it is first necessary to understand pulmonary ventilation and perfusion in healthy lungs. Pulmonary perfusion has not been studied as intensely as ventilation even though it is indicated that lung volume greatly affects perfusion [4].

Previously Fung and Sobin [5] have modelled the pulmonary capillary perfusion as a sheet flow through a two-dimensional network built on morphometric studies on cat lungs. Burrowes et al. [6] used an anatomically based finite element model to simulate regional variations in blood perfusion. Burrowes and Tawhai [7] constructed an arterial geometric model using a combination of computed tomography and a volume-filling branching algorithm. In addition, Liu et al. [8] modelled the airway mechanics, gas exchange, and perfusion in a nonlinear model excluding the hydrostatic effects of tissue and blood. None of these previous models of the pulmonary capillary perfusion have described how the perfusion is affected by the airway pressure and lung volume at different lung heights. Mogensen et al. [9] used a physiological stratified model to simulate the influence of different lung volumes on perfusion. The model did, however, only take the lung volume as input and did not include the hydrostatic effects of changing lung height and density.

This paper presents a modified version of the physiological model by Mogensen et al. [9], which describes the pulmonary microcirculation, enabling simulation of capillary

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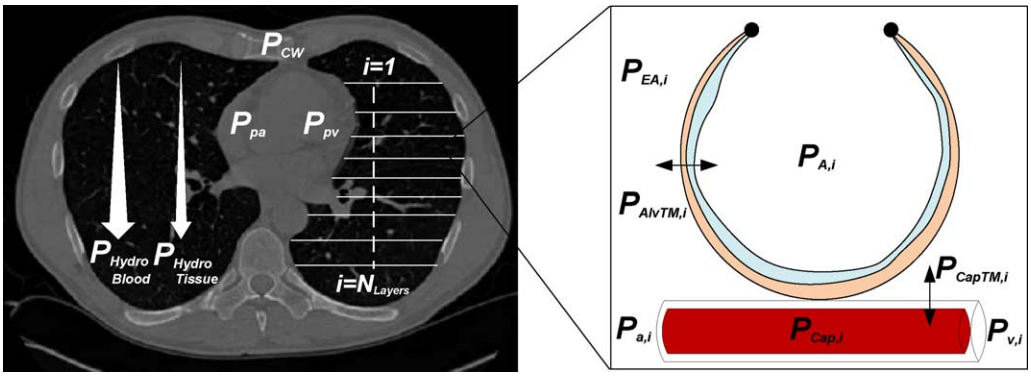

Fig. 1 – Illustration of the total perfusion model. P_{EA}: extraalveolar pressure; P_A: airway pressure; P_{Cap}: capillary pressure; P_{CapTM} and P_{AlvTM}: capillary and alveolar transmural pressures; P_a and P_v: arterial and venous blood pressure; P_{pa}: pulmonary artery pressure at the pulmonary valve; P_{CW}: pressure exerted by the chest wall; $P_{HydroTissue}$ and $P_{HydroBlood}$: hydrostatic pressure due to lung tissue and blood; i: index controlling layer number measured from the top.

blood perfusion around the alveoli in the entire lung. The model describes the lungs as divided into a number of horizontal layers as defined by gravity. The model includes aspects of the capillary geometry, hemodynamics and blood rheology. The model includes the extraalveolar pressure which depends on lung height, density and the total lung volume, which change during breathing. Model simulations are compared to measurements of the perfusion distribution in the entire pulmonary microcirculation during mechanical ventilation; the total capillary blood perfusion; capillary blood volume; capillary surface area and transition time during different ventilator settings.

2. Methods

2.1. Model introduction

The model has been implemented in MatLab (Mathworks, Natick, MA). The model simulates a healthy human lung at rest during mechanical ventilation. Fig. 1 illustrates the model representation of the pulmonary system. The lungs are modelled by 100 layers each reflecting a vertical height. For a person in the supine position the layers are in the frontal plane and numbered in the vertical–dorsal direction. In Fig. 1 only 10 layers are depicted for the sake of simplicity. The lung height (non-dependent to dependent) must be included in the model because the lung tissue weighs down on the layers below causing a hydrostatic gradient, $P_{HydroTissue}$. The blood in the capillary network also imposes a hydrostatic gradient that increases the blood pressure down the lungs, $P_{HydroBlood}$. The hydrostatic pressure of blood is calculated with reference to the pressure in the pulmonary valve at position D_0, which is assumed to be 5 cm down the lungs in supine position [10]. The hydrostatic pressures due to blood and tissue are calculated by Eq. (1) and Eq. (2), respectively.

$$P_{HydroBlood,i} = \sum_{j=1}^{i} \rho_{Blood} \cdot g \cdot (t_j - D_0) \quad (1)$$

$$P_{HydroTissue,i} = \sum_{j=1}^{i} \rho_{Lung,j} \cdot g \cdot t_j \quad (2)$$

where ρ_{Blood} is the density of the blood, assumed to be constant, g is the gravitational acceleration and t_j and $\rho_{Lung,j}$ are the thickness and lung tissue density of layer j, respectively. Both density and thickness of the different layers change during breathing. In order to calculate the density and thickness of each layer, the volume of air in each layer must be known. This causes a problem since the input to the model is the airway pressure or alveolar pressure, P_A, and the distribution of air between alveoli at different layers at a given input pressure is dependent on the densities and the thicknesses of the layers. To solve this problem, a model of the pulmonary ventilation is used [11]. This stratified model simulates the pulmonary alveolar ventilation in layers identical to those in the model presented here. A limitation in the ventilation model is that it assumes that during normal tidal breathing at rest the effect of airway resistance and viscoelasticity are minimal. The ventilation model uses P_A as input to calculate densities and thicknesses of the different layers. It, however, does not divide the pulmonary blood into capillary blood and blood in larger vessels [11]. The formula for lung density, ρ_{Lung}, presented by Steimle et al. [11] is therefore expanded to Eq. (3). It is assumed that air density is negligible.

$$\rho_{Lung,i} = \frac{\rho_{Tissue} \cdot V_{TissuePrLayer} + \rho_{Blood} \cdot (V_{CapBlood,i} + V_{VesselBloodPrLayer})}{V_{TissuePrLayer} + V_{CapBlood,i} + V_{VesselBloodPrLayer} + V_{Air,i}} \quad (3)$$

where $V_{CapBlood,i}$ and $V_{Air,i}$ are the volumes of capillary blood and air in layer i. $V_{TissuePrLayer}$ and $V_{VesselBloodPrLayer}$ are tissue and blood in larger vessels per layer. ρ_{Tissue} and ρ_{Blood} are densities of tissue and blood, both assumed constant.

$V_{Air,i}$ is determined using the model by Steimle et al. [11]. It is assumed that the blood in larger vessels and tissue is equally distributed between the layers. The volumes of the lung tissue per layer, $V_{TissuePrLayer}$, and blood in larger vessels