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PERSONALIZED PULMONARY POROMECHANICS IN HEALTH, IDIOPATHIC PULMONARY FIBROSIS AND COVID-19

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SUMMARY

This abstract briefly describes our recent lung poromechanical model, and its associated model personalization procedure based on routine clinical data (thoracic scanners at end-expiration and end-inspiration). The final aim of the personalized modeling pipeline is to provide automatic and objective diagnosis tools for fibrosing interstitial lung diseases, in the form of estimated regional stiffness maps. Here we present early results on three idiopathic pulmonary fibrosis patients and one COVID-19 patient. In the presentation we will detail the impact of various parameters of the method.

Key words: *pulmonary mechanics, poromechanics, modeling, estimation*

1 INTRODUCTION

Lung biomechanics has been extensively studied by physiologists, experimentally as well as theoretically, laying the ground for our current fundamental understanding of the relationship between function and mechanical behavior. However, many questions remain, notably in the intricate coupling between the multiple constituents. These fundamental questions represent real clinical challenges, as pulmonary diseases are an important health burden. Interstitial lung diseases, for instance, affect several million people globally [1]. Idiopathic Pulmonary Fibrosis (IPF), notably, a progressive form of fibrosing interstitial lung diseases where some alveolar septa get thicker and stiffer while others get completely damaged, remains poorly understood, poorly diagnosed, and poorly treated [2]. Furthermore, fibrosing interstitial lung diseases are a potential chronic complication of COVID-19 [3].

2 METHODS

2.1 Lung poromechanical model

We recently developed a model of the lungs at the breathing time scale and the organ space scale [4, 5], based on a general poromechanical formulation compatible with large strains and thermodynamics [6], where the “solid” phase is composed of both tissue and blood while the fluid phase is the air. Several pulmonary-specific hypotheses have been formulated, assuming that the transformation is quasi-static and that the fluid pressure inside the alveoli is homogeneous. The problem is then simplified and becomes a coupled problem between the two unknowns, the (vector) displacement \underline{U} and the (scalar) porosity Φ .

In the model, specific boundary conditions are imposed on the lungs themselves, modeling the effect of diaphragm-induced loading and rib cage: a pressure applied on lung surface representing pleural

pressure, and a frictionless contact with the moving thorax [5]. Moreover, the constitutive behavior of our model allows to reproduce the volumetric response of lungs to a change of pressure as observed in experimental data, as well as the quasi-incompressibility of the solid phase [5].

2.2 Model personalization

The proposed model can be personalized using clinical data [7, 8]. Two 3D CT-scans I_0 and I_1 , acquired at end-exhalation and at end-inhalation respectively, are used (i) to generate (after segmentation [9]) a patient-specific geometry (see Figure 1), (ii) to get a personalized porosity field, and (iii) to personalize parts of the boundary conditions after motion tracking [10].

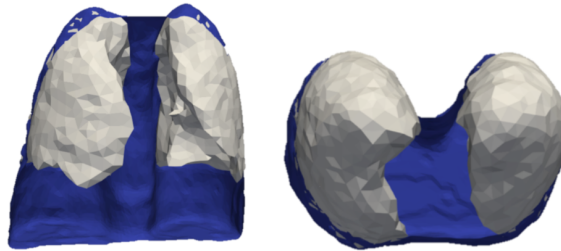


Figure 1: Geometrical model of the lungs and thorax, obtained from biomedical images [7].

In principle, pleural pressure could be measured in patients and used in the model personalization procedure. It is, however, a very invasive procedure, not part of clinical routine. As a consequence, in this work we either prescribe a normal pressure value, or estimate it.

The data allows also to personalize parts of the model. Regional mechanical parameters are estimated by minimizing the error between data and model. We investigated two different cost functions, differing by the nature of the data considered: either a FEMU strategy is used, considering lung volume displacement computed by image registration [10], or an integrated image correlation approach is used, based on the 3DCT images directly. We also investigated two types of parametrization of the problem, where the estimated material parameters can be the absolute (i.e., independent from porosity, in which case the effective parameters depend linearly on the porosity) or the effective (i.e., those characterizing the mixture) parameters. Finally, we also considered the following two parametrizations: either the two unknowns are material stiffness in healthy and diseased regions (in which case the applied pressure is fixed), or they are the diseased stiffness and the applied pressure (in which case the healthy stiffness is fixed), to take into account the fact that breathing pressure is more uncertain than healthy lung tissue properties.

Since the images correspond to a loaded configuration, we also focused on the inverse problem of finding the unloaded configuration associated with the loaded configuration observed in vivo, and corresponding issues in the poromechanical setting. Indeed, special care should be taken to ensure a positive porosity for the unloaded configuration. We proposed two methods to that purpose, using either an additional energy describing the collapse of the mixture pores or a contact-like formulation.

3 RESULTS

Using images from patient suffering from pulmonary diseases, regional mechanical parameters and/or breathing pressures can be estimated in diseased lungs. The personalization was applied to a control subject (patient without pulmonary disease and normal thoracic scan), three patients with IPF, and one patient with COVID. For IPF patients, we found that the model becomes closer to the data in the case where two material regions are defined, corresponding to the healthy and the diseased regions of the lung, as observed on the images. The same trend appears when the problem is parametrized using effective or absolute parameters.

Results are presented for the three IPF patients and the COVID patient, see Figure 2. They are based on the integrated image correlation approach, and the absolute parametrization of the compliance. In the figure we compare the estimation parametrizations with fixed end-inspiration pleural pressure

versus fixed healthy compliance. In all cases, we found that the diseased region results in a stiffer material than the healthy region, which is consistent with the fact that fibrosis leads to local tissue stiffening [2].

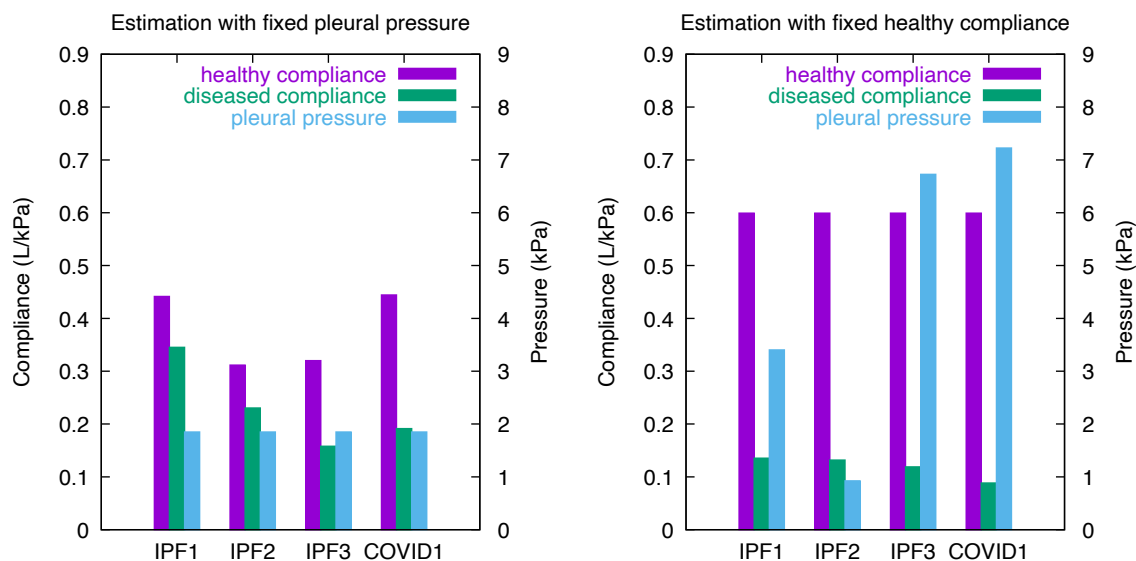


Figure 2: Compliances and/or pressure estimated for three IPF and one COVID patients. Left: A normal end-inspiration pressure of 1.85 kPa is prescribed, and both healthy & diseased compliances are estimated. Right: A normal healthy compliance of 0.6 L/kPa is prescribed, and both the diseased compliance and end-inspiration pressure are estimated.

4 CONCLUSIONS

Our results illustrate how our pulmonary poromechanical model could be used as an augmented, i.e., objective & quantitative, diagnosis tool in the clinic when it is personalized to a patient using clinical data acquired in routine.

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