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Fluid Movement in Porous Bone via Blood Pressure: A Porous Media Theory

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Abstract **-** It is widely believed that fluid flow through the network of osteocyte canaliculi is the primary factor that controls cortical bone adaptation. Due to the difficulty of considering mass exchange between different porosity sizes in cortical bone and the measurement of the influence of the blood pulsatile on interstitial fluid using poroelasticity theory, the theory of porous media is an attractive alternative for studying cortical bones. This study presents a dual porosity model with the theory of porous media including mass exchange between vascular and lacunar-canalicular porosities including blood pressure variations on the interstitial fluid. This model enables measuring and analyzing interstitial fluid's velocity in the vascular porosity (PV) and interstitial fluid's pressure in the lacunar-canalicular porosity (PLC). The results without considering blood pressure variations verify that the predicted fluid flow field follows a general pattern consistent with that obtained from earlier studies. Taking blood pressure pulses into consideration changes the velocity and pressure fields of the interstitial fluid within the cortical bone. In addition, this method has the potential to consider mass exchange between the solid and fluid phases in relation to chemical reactions within the cortical bone.

Keywords**:** Blood pressure, Mass exchange, Bone dual porosity, Kozeny–Carman equation, Bone remodeling

1. Introduction

Cortical and cancellous bones comprise bone that is a functionally graded material. Cancellous bone is a lattice of large plates and rods called trabecula, while cortical bone is a dense, solid mass [1]. Cortical and cancellous bone are made up of a hierarchical network of flow channels with varied characteristic diameters.

There are three types of porosity in cortical bone [2]: vascular porosity (PV), lacunar-canalicular porosity (PLC), and collagen-apatite porosity (PCA). Vascular porosity (PV) is comprised of all quasi-cylindrical passages in the bone matrix, such as osteonal canals and Volkmann canals, that contain blood vessels, nerves, and bone fluid. PV has a characteristic radius of approximately 20 μm; Lacunar-canalicular porosity (PLC) is characterized by spaces in the lacunae and canaliculi with a radius of approximately 0.1 μm; Collagen-apatite porosity (PCA) associated with the spaces between the collagen and the crystallites of the mineral apatite. PCA has a characteristic radius of approximately 10 nm, so there is negligible movement of bone fluid in PCA. Bone fluid is the fluid contained within the porosities of the bone, but outside the blood vessels and nerves. In relation to physiological factors, the structure of the bone and the shape of the porosities change over time.

Bone can adapt its structure to its mechanical environment in order to provide the optimal shape with the least amount of material [3]. Bone will attain its optimal shape through two interconnected processes: growth and remodeling. The remodeling process modifies the bone structure, whereas growth refers to the accretive or resorptive variation in bone mass [4]. While the biological process of bone adaptation has been well-established, the precise mechanical stimulus that controls it remains unclear. To investigate mechanical stimuli that activate bone, it is necessary to first discuss bone cells.

The bone contains four distinct types of cells: osteocytes, osteoblasts, osteoclasts, and lining cells. Osteoclasts are responsible for bone resorption, whereas osteoblasts are responsible for bone production. After bone formation, osteoblasts are buried and transformed into osteocytes within the bone. Osteocytes are mechanosensitive bone cells that respond to mechanical loadings. Bone lining cells, on the other hand, are at rest on the bone's surface and transform into osteoblasts in response to bone injury or irregular loading to initiate bone healing or remodeling [5]. The activation of each of these cells within the bone in response to physiological stimuli is still unknown, so numerous research have been studied to explain the process of bone remodeling.

Recent studies by [6] and [7] have proposed that pressure and velocity of the interstitial fluid is also important during bone remodeling in addition to strain and stress of the solid part. It has been demonstrated that fluid velocity and shear strain

of the velocity field are stimuli for "Mesenchymal Stem Cells" (MSC) activities in [6]. Osteoblasts and osteocytes are derived from pluripotent MSCs, as demonstrated by [8]. A significant correlation between octahedral shear strain and tissue type differentiation, particularly in the early stages of bone healing is reported in [9]. It has been demonstrated in [10] and [11] that mechanical loading and its corresponding small portion of solid strain field, which is between 0.2 and 0.3 percent for physiological loads discussed in [12] and [13], cause fluid flow circulation in the canalicular network. Through this circulation, the fluid component is forced through the osteocyte canaliculi, creating a shear stress field that stimulates osteocytes to produce signaling molecules that recruit osteoblasts and osteoclasts [14] and [15]. Consequently, it is more accurate to model bone remodeling by considering bone as a bi-phasic mixture consisting of a solid and an interstitial fluid phase.

Regarding the various porosities of the cortical bone and the movement of interstitial fluid between the levels of porosity, it is necessary to employ a method for capturing mass exchange between levels of porosity to study the process of bone remodeling. With the extended poroelasticity theory [16] for dual porosity mixtures, the displacement of the solid phase and pressure of the interstitial fluid in PV and PLC is studied in [17]. Study by [17] disregards two crucial variables that could affect the displacement and pressure and velocity fields of solids and fluids: mass exchange between the solid and fluid phases and arterial pressure inside the PV. Due to the inefficiency of Biot's poroelasticity theory to account for mass exchange between phases, a different method must be used to consider these two parameters.

Based on the formulation for the theory of porous media, the current work develops a generalized treatment for determining the solid's displacement, the interstitial fluid's pressure in PLC, and the interstitial fluid's velocity in PV. Furthermore, the advection term that is the change in the momentum by advection is considered in this study, while previous investigations disregard this term. In calculating interstitial fluid fields, time-varying blood pressure in the PV will be accounted for (Fig. 1 (a)). In addition, mass transfer between the solid and fluid phases will be investigated. For solving the fluid part equations and meshing the model Ansys Fluent is used. Afterwards, the Fluent results of each node are imported into a Matlab code that calculates the displacement of the solid part.

2. Results and Discussion

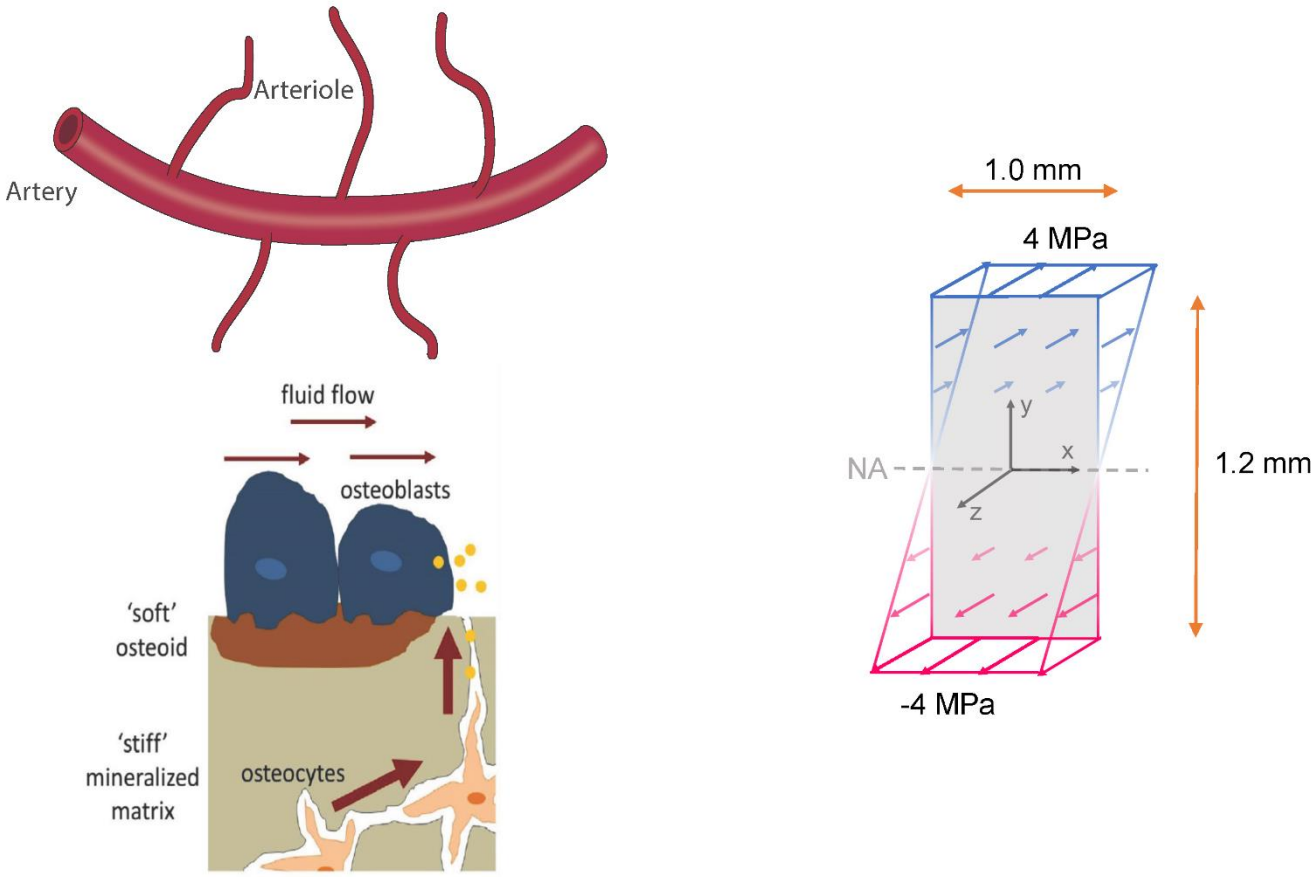
A saturated mixture with dimensions of 1.0 mm in width and 1.2 mm in height is considered in a transient 2D problem (Fig. 1 (b)). The lacunar-canalicular porosity was assumed to be five percent of the medium, while the vascular porosity was modeled with three different values: 0.02, 0.05, and 0.07; both porosities are filled with interstitial fluid. The elastic constants of cortical bone in both solid and fluid phases are listed in Table 1.

Table 1: Material constants of solid and fluid phases of the cortical bone in section 2.

Materials	ρ (Kg/m^3)	E (GPa)	ν	K (GPa)
Cortical bone [a]	4.985×10^{5} [b]	17	0.232	16
Interstitial fluid [c]	1000	-	-	2.3

[a]According to [18]; [b]According to [1], and [c]According to [17].

The shear stress of the fluid component is ignored in this problem. At a temperature of 20 ˚C, the viscosity of water is assumed to be 1.002 mPa.s. The leakage coefficient and PV permeability are calculated using Fig. 2. According to [19] and [17] investigation, a logarithmic equation regarding PV porosity is considered for the leakage coefficient, as depicted in Fig. 2. The selection of a logarithmic equation for the leakage coefficient is because increasing the PV porosity diminishes the effect of the mass exchange between PLC and PV. In addition, Kozeny–Carman equation [20], [21], and [22], which relates the pressure drop of a fluid to the porosity and permeability of the solid component in a mixture, is assumed for a relationship between the porosity and permeability of the PV. This equation is valid for laminar flows, so its application to the flow of interstitial fluid within the bone is appropriate.



(a) (b)

Fig. 1: (a) The biomechanical environment experienced by osteoblasts and osteocytes regarding flow between PV and PLC with considering an artery adapted from [18] and modified, (b) shape of the proposed problem.

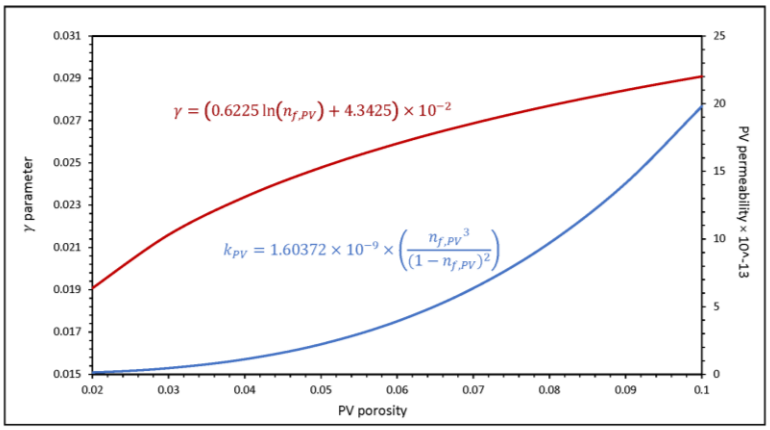


Fig. 2: Leakage coefficient and PV permeability for PV porosity according to [19] and [17].

For one second, a bending moment is considered with the mixture's top and bottom boundary stresses of 4 MPa and -4 MPa, respectively, while all other boundaries are considered impermeable (Fig. 1 (b)). Meanwhile, blood pressure is thought to beat in the pattern depicted in Fig. 3, as proposed by [20].

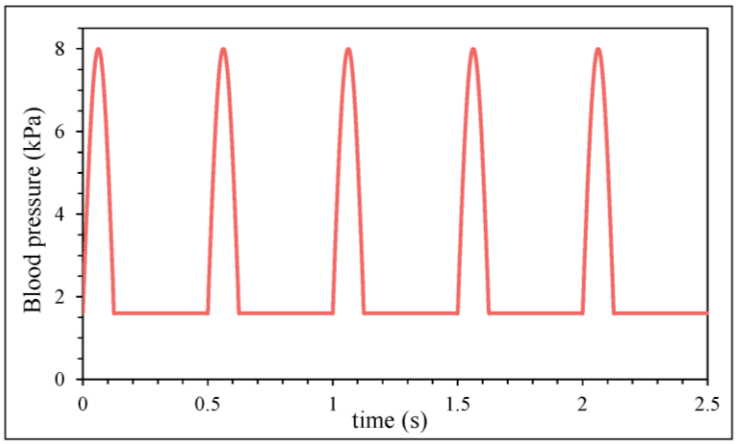


Fig. 3: Pulsatile flow profile considering blood pressure variation instead of blood velocity variation during time [20].

Figs. 4 (a) and 4 (b) illustrate lacunar-canalicular pressure variations at a point with coordinates of (0, -600) μm over 0.0-1.5 seconds and 1.0-1.5 seconds, respectively. As shown in Fig. 4 (a), larger vascular porosity sizes reduce PLC pressure because more interstitial fluid in the PLC can diffuse inside the PV. Furthermore, Fig. 4 (a) illustrates the pressure drop between porosities decreases with increasing vascular porosity. Fig. 4 (b) depicts lacunar-canalicular pressure variations after loading considering only blood pressure. Smaller vascular porosities result in lower interstitial fluid pressure due to PV's blood pressure inside the lacunar-canalicular porosity. As shown in Fig. 4 (b), pressure variation between different porosities is greatest for smaller PV porosities and decreases as PV increases.

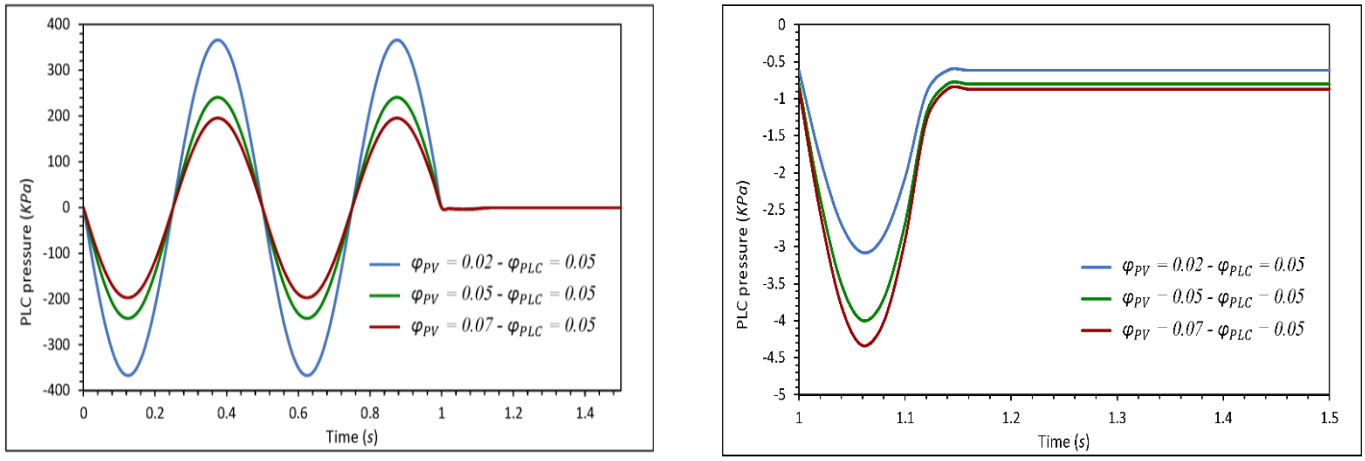


Fig. 4: Lacunar-canalicular pressure variations at the point (0, -600) μm through: (a) 0.0-1.5 seconds and (b) 0.0-1.5 seconds.

Figs. 5 depicts interstitial fluid velocity variations in vascular porosity at the point with coordinates of (0, -600) μm over 0.0-1.5 seconds. Even though the order of velocity magnitude in Fig. 6 is 10^{-6}, velocity is crucial to the bone

remodeling process. Through the vascular porosity, the velocity of interstitial fluid exerts a shear stress on the lining cells (Fig. 1 (a)). As expected, velocity at point (0, -600) μm decreases as vascular porosity increases during the loading time (0.0-1.0 seconds) as depicted in Fig. 6. Greater vascular porosity dissipates the applied pressure more effectively, resulting in a slower velocity for the interstitial fluid. Considering interaction term between solid and fluid phases decrease the velocity of the PV drastically at the point (0, -600) μm. This change in velocity variations considering the interaction term is because the movement of interstitial fluid in the PV becomes increasingly difficult from one point to another. Interstitial fluid movement is dependent on the permeability of the bone and the viscosity of the interstitial fluid, which they are related to each other by the interaction term between the solid and fluid phases.

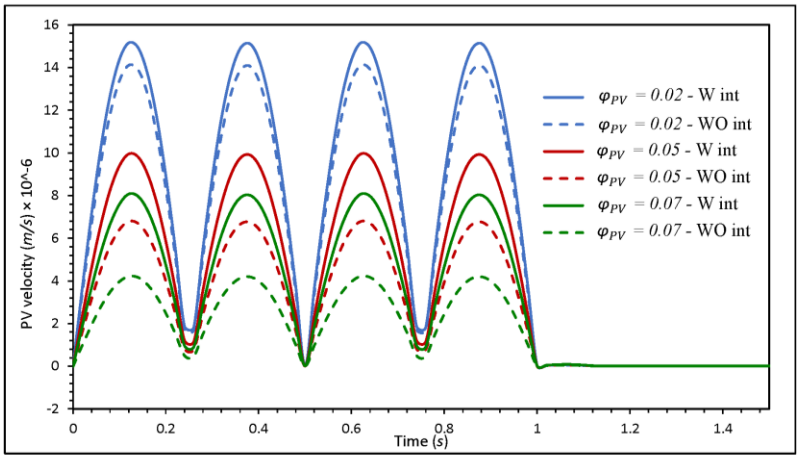


Fig. 5: interstitial fluid's velocity variations in PV through 0.0-1.5 seconds at the point (0, -600) μm with and without the interaction term between the solid and fluid phases.

3. Conclusion

This study investigates the interstitial fluid's pressure in PLC and velocity in PV considering blood pressure variation, loading on cortical bone, and interstitial fluid movement between PV and PLC with the theory of porous media. Without taking into account blood pressure variations, the results show that the predicted fluid flow field follows a general pattern consistent with previous research. While considering blood pressure variation results in changes to the pressure and velocity fields. The current formulation allows for the consideration of chemical reactions between the solid phase of cortical bone and the interstitial fluid, which merits further investigation.

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