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Simulating vertebroplasty: A biomechanical challenge

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Every year, 1.4 million osteoporosis-related vertebral fractures occur worldwide, which cause about 33 % of all osteoporotic fracture-related deaths. Vertebroplasty is a medical procedure for treating vertebral fractures, wherein a polymer called bone-cement is injected into the porous structure inside the vertebra. The associated risk of bone-cement leaking outside of the vertebra can be mitigated by simulating the procedure, which can help clinicians to determine the right operating parameters in advance. However, modelling this procedure is challenging as well as computationally expensive using conventional methods, due to factors like the complex geometry of the trabecular bone structure, non-Newtonian rheology of the fluids, curing of bone-cement, and unknown patient-specific material properties. In this work, some of these challenges are addressed by using a multiphasic continuum approach and adapting the constitutive models accordingly. The resulting developed model provides a computationally feasible framework for simulating the spread of bone-cement in the whole vertebra.

1 Introduction

The main challenges in simulating vertebroplasty arise due to the facts that: (i) the trabecular structure is geometrically complex, making it computationally too expensive to simulate the flow through the whole vertebra with direct numerical simulations; (ii) the bone-cement and the bone marrow behave both as non-Newtonian fluids, i.e. their viscosities vary with the shear rate; (iii) the bone-cement undergoes curing while it is being injected; and (iv) the properties of the biomaterials and the conditions are often unknown and vary from case to case. In this work, some of these challenges are addressed using a multiphasic continuum model based on the Theory of Porous Media [**?**]. The homogenised continuum approach makes a model computationally cheap to simulate the flow through the trabecular structure. In this regard, the Darcy law (used to compute the filter velocity at the macro-scale) is made applicable for the non-Newtonian rheologies by upscaling the shear rate using the so-called Cannella model. The resulting model represents a computationally feasible macro-scale model for bone-cement flow through the whole vertebra that also incorporates the effect of pore-scale geometry and material behaviours.

2 Modelling framework and constitutive setting

The domain consists of three immiscible phases: the solid trabecular bone φ^S, the bone marrow φ^M, and the bone-cement φ^C. These are homogenised over the whole domain leading to superimposed but interacting continua. The aggregate is given by $\varphi = \cup \varphi^\alpha$ with $\alpha = \{S, C, M\}$. The individual local amount is considered via volume fractions n^α, such that $\sum_\alpha n^\alpha = 1$. The motion of the solid φ^S is described in a Langrangean setting via the displacement $\mathbf{u}_S$, while that of the liquids is described in a modified Eulerian setting via seepage velocities $\mathbf{w}_\beta$, with $\beta = \{C, M\}$. The balance relations are obtained after assuming materially incompressible phases, no mass exchange, quasi-static and isothermal processes, and uniform body forces $\mathbf{b}$ as

$$n^S = n^S_{0S}(\det \mathbf{F}_S)^{-1}, \quad (n^\beta)'_S + \operatorname{div}(n^\beta \mathbf{w}_\beta) + n^\beta \operatorname{div} \overset{\prime}{\mathbf{x}}_S = 0, \quad \text{and} \quad \operatorname{div} \mathbf{T} - \rho \mathbf{b} = 0. \tag{1}$$

Additionally, the Darcy filter velocities $n^\beta \mathbf{w}_\beta$ are obtained as

$$n^C \mathbf{w}_C = -\frac{\kappa^C_r \mathbf{K}^S}{\mu^{CR}}(p^{CR} - \rho^{CR}\mathbf{b}) \quad \text{and} \quad n^M \mathbf{w}_M = -\frac{\kappa^M_r \mathbf{K}^S}{\mu^{MR}}[p^{MR} - \rho^{MR}\mathbf{b} - \frac{p_{\text{dif.}}}{n^M}(s^C \operatorname{grad} n^M - s^M \operatorname{grad} n^C)]. \tag{2}$$

where $\mathbf{F}_S = \mathrm{d}\mathbf{x}/\mathrm{d}\mathbf{X}_S$, $\mathbf{T} = \mathbf{T}^S_E - p\mathbf{I}$ is the stress tensor, $p = s^C p^{CR} + s^M p^{MR}$ is the overall pore pressure with s^β being the saturations and $p^{\beta R}$ the fluid pressures, $p_{\text{diff.}} = p^{CR} - p^{MR}$, κ^β_R and $\mathbf{K}^S$ are permeabilities, and $\mu^{\beta R}$ are the viscosities.

The constitutive relations include the linear elastic law for the solid extra stress $\mathbf{T}^S_E$, the van Genuchten relation for the saturations $s^M_{\text{eff}} = [1 + (\alpha_{\text{gen}} p_{\text{dif.}})^{j_{\text{gen}}}]^{-h_{\text{gen}}}$, and the relative permeabilities $\kappa^C_r = -0.9 s^M_{\text{eff}} + 1.0$, $\kappa^M_r = 0.9 s^M_{\text{eff}} + 0.1$.

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Table 1: The used material parameters are extracted from the literature. The van Genuchten parameters are estimated. $\mathbf{K}^S = K^S\mathbf{I}$ for the isotropic permeability. The parameters for bone-cement and bone marrow are fitted from measured data from [**?**] and [**?**] respectively.

Parameter	Value	Parameter	Value	Parameter	Value	Parameter	Value
μ^S	$7.7 \cdot 10^7$ N/m^2	ρ^{CR}	1100 kg/m^3	k_3	$3.7 \cdot 10^{-9}$ Pa s	μ_∞ (marrow)	5.0
λ^S	$1.1 \cdot 10^8$ N/m^2	K^S	$5 \cdot 10^{-8}$ m^2	k_4	42.83 Pa s	α (marrow)	9.0
n^S_{0S}	0.15	α_{gen}	$1 \cdot 10^{-4}$ Pa^{-1}	C	6.0	λ (marrow)	38.37 s
n^M_{0S}	0.80	j_{gen}	3.2	α (cement)	2.38	n (marrow)	0
n^C_{0S}	0.05	h_{gen}	0.6875	n (cement)	0.5584		
ρ^{SR}	2720 kg/m^3	k_1	1265.4 Pa s	λ (cement)	5.263 s		
ρ^{MR}	1000 kg/m^3	k_2	-2367.0 Pa s	μ_0 (marrow)	3199 Pa s		

Therein, $s^M_{\text{eff}} = (s^M - s^M_{\text{res}})/[1 - s^M_{\text{res}} - s^C_{\text{res}}]$, s^C_{res} and s^M_{res} are residual saturations, and α_{gen}, j_{gen}, h_{gen} are van Genuchten parameters. The viscosities are obtained using the Carreau model, and the shear rate $\dot\gamma_{\text{eff}}$ is obtained using the Cannella model.

$$\mu^{\beta R} = \mu_\infty + \frac{\mu_0 - \mu_\infty}{[1 + (\lambda\,\dot\gamma_{\text{eff}})^\alpha]^{\frac{1-n}{\alpha}}} \quad \text{with} \quad \dot\gamma_{\text{eff}} = C\left[\frac{3n+1}{4n}\right]^{\frac{n}{n-1}}\left[4|\mathbf{w}_\beta|\sqrt{\frac{n^\beta}{8\kappa_r^\beta K^S}}\right] \tag{3}$$

K, n, λ, α, C are material parameters. μ_0 and μ_∞ are asymptotic viscosities: For the curing bone-cement, they are written as $\mu_0 = k_1(t/t_0) + k_2$ and $\mu_\infty = k_3(t/t_0) + k_4$. t is the curing time, $t_0 = 1.0$ s, and $k_{1..4}$ are fitting parameters.

3 Numerical simulation

The geometry of the vertebra was extracted from CT images provided by the AO Foundation (Davos). It was then modified to replicate the insertion of a 2.4 mm diameter needle and meshed using tetrahedral elements. The boundary conditions and the material parameters are as shown in Figure **??** and Table **??** respectively. The injection starts at 3 minutes of curing time. The model was implemented and solved using the coupled FE solver PANDAS. The results show the spread of bone-cement with time. The required injection pressure increases with time due to the curing of the bone-cement.

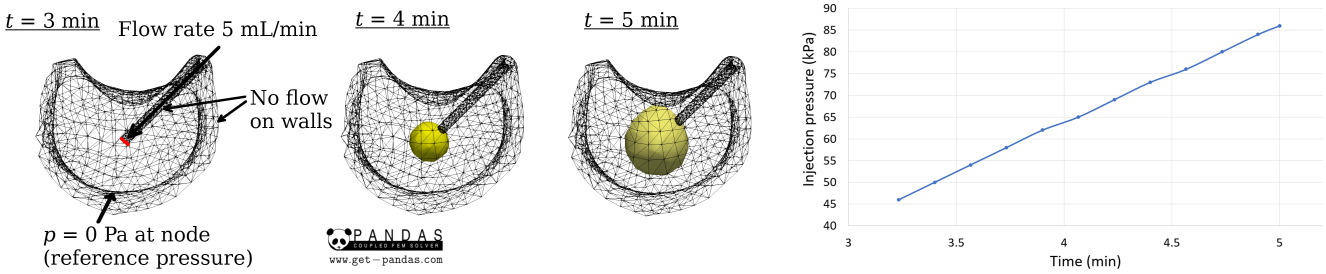

Fig. 1: Boundary conditions and bone-cement spreading with time (left), increase of injection pressure over time (right).

4 Conclusion and outlook

The current work presents a multiphasic continuum modelling framework as a computationally feasible approach for simulating vertebroplasty in a pre-clinical study, while also including the complex rheologies of the fluids. However, the Cannella model used for upscaling the shear rate is a semi-empirical approach. Apart from this, determining the material parameters remains a challenge, e.g. the van Genuchten parameters. The flow behaviour is mostly sensitive to the permeability. A sound patient-specific estimation thereof is one of the biggest challenges. These limitations provide the ground for future work. In addition, experiments are needed for validating the model.

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