3-D Analyses of Changes in Free Bubble induced Stresses on Blood Vessel Wall in Ultrasound Therapy

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INTRODUCTION: A microbubble inside a blood vessel in the presence of ultrasound waves induces stresses on the vessel wall [1]. We simulated the changes in the induced circumferential and shear stresses on the vessel wall for a bubble drifting away from the vessel axis under ultrasound pressure of 150 kPa at 1 MHz. The 1.5-µm radius bubble was placed in 6 different positions (0, 0.5, 1, 1.5, 2, 2.5), starting from on the vessel axis to 2.5 µm away from axis in a 5-µm radius vessel.

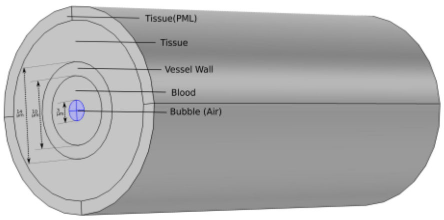

Figure 1. A 3-D bubble-blood-vessel-tissue model with bubble placed on the vessel axis.

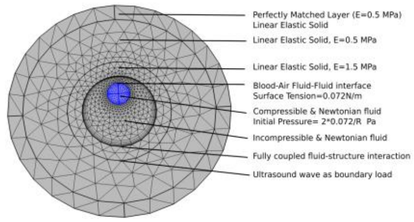

Figure 2. A meshed 2-D view of model with bubble placed 2.5 µm away from vessel axis labelled with necessary physical parameters.

Variable	Blood	Air
Density	1055 kg/m^3	Initial=2.3843 kg/m^3 (changes with pressure)
Dynamic Viscosity	0.005 Pa-s	1.814e-5 Pa-s

Table 1. Material Properties of fluid domain

Variable	Vessel	Tissue
Density	1070 kg/m^3	1050 kg/m^3
Young's Modulus	1.5 MPa	0.5 MPa
Poisson's Ratio	0.49	0.49

Table 2. Material Properties of solid domain

METHODS:

- FLUID DOMAIN: Blood and bubble (air) domains were solved using Navier-Stokes and continuity equation with blood as Newtonian & incompressible fluid and bubble as Newtonian & compressible fluid. The bubble and blood boundary was given fluid-fluid interface with surface tension of 0.072 N/m. The bubble was given Laplace Pressure equivalent to 2*0.072/R Pa to balance the surface tension. (R=bubble radius)

$$\rho\frac{\partial \boldsymbol{v}}{\partial t}+\rho(\boldsymbol{v}\cdot\nabla)\boldsymbol{v}=\nabla\cdot\left[-pI+\mu(\nabla\boldsymbol{v}+(\nabla\boldsymbol{v})^{T})-\frac{2}{3}\mu(\nabla\cdot\boldsymbol{v})I\right]+\boldsymbol{F},\qquad \textit{where } \rho \textit{ is fluid density}, v \textit{ is velocity vector}, p \textit{ is pressure},$$

$$\frac{\partial\rho}{\partial t}+\nabla\cdot(\rho\boldsymbol{v})=0\qquad \mu \textit{ is fluid viscosity and } F \textit{ is volume force vector}$$

- SOLID DOMAIN: The vessel and tissue were assumed as linear elastic solid and deformation was solved using below equation. A perfectly matched layer (PML) was given to outer boundary of tissue to absorb the oscillations and prevent reflections.

$$\rho_s\frac{\partial^2\boldsymbol{u}}{\partial t^2}=\nabla.\sigma+\boldsymbol{F_v},\qquad \textit{where } \rho_s \textit{ is solid density}, u \textit{ is displacement vector}, \sigma \textit{ is stress tensor and } F_v \textit{ is volume force vector}$$

- MESH: Tetrahedral element were used for meshing with linear interpolation for velocity and pressure in fluid domain and quadratic Lagrange interpolation for displacement in solid domain. An ALE (Arbitrary Lagrangian Eulerian) based moving mesh was used for fluid domain to account for deformation at vessel-blood interface and bubble-blood interface. The total mesh have around 45000 elements with very fine mesh on bubble-blood interface and boundary layer mesh on blood-vessel interface.
- BOUNDARY CONDITIONS & TIME STEP: The ends of the solid and fluid domain were given symmetry boundary condition to assume the model as infinitely long. The ultrasound wave was applied to the outer side of vessel using the boundary load condition. A fully coupled fluid structure interaction along with no slip was applied at blood-vessel interface. A free time step with maximum time step of 3 nanosecond was used for all cases. The model was solved till 4.5 µs.
- STRESSES CALCULATION: The circumferential stress (CS) on the vessel wall due to pressure exerted by blood was calculated by assuming the vessel as a thick cylinder using below equation. The shear stress(SS) on the vessel wall due to blood velocity gradient near the wall is calculated using below equation.

$$(CS)\quad \sigma_{cr}=\frac{P_i r_i^2}{r_o^2-r_i^2}+\frac{r_i^2 r_o^2}{r^2}\left(\frac{P_i}{r_o^2-r_i^2}\right),\qquad \textit{where Pi is the pressure on the inner side of vessel, } r_i \textit{ and } r_o \textit{are the inner and outer vessel radius}$$

$$(SS)\quad \tau_{xz}=\mu\left(\frac{\partial u}{\partial z}+\frac{\partial w}{\partial x}\right),\qquad \text{where } \mu \textit{ is the blood viscosity}, u \textit{ is the blood velocity in x direction and } w \textit{ is the blood velocity in z direction}$$

- VERIFICATION: The model was verified by solving the model with large vessel radius (10 times of bubble radius) to reduce the vessel confinement effect and comparing the obtained results with Keller-Miksis equation's results.

RESULTS:

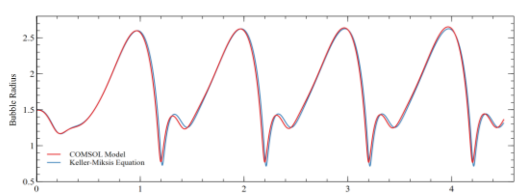

Figure 3. Model verification by comparing with Keller-Miksis equation results.

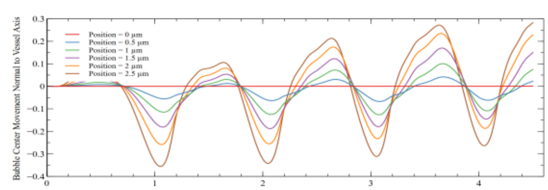

Figure 4. Bubble center movement in the direction perpendicular to the vessel axis direction.

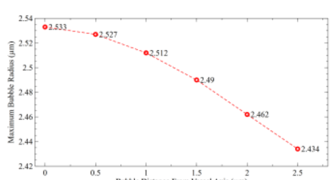

Figure 5. Maximum bubble radius as a function of bubble distance from vessel axis.

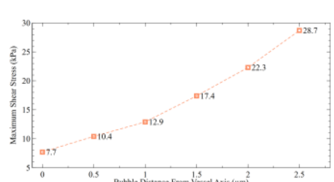

Figure 6. Maximum shear stress as a function of bubble distance from vessel axis.

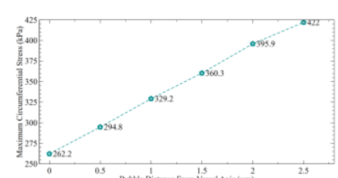

Figure 7. Maximum circumferential stress as a function of bubble distance from vessel axis.

CONCLUSIONS: The bubble oscillation for larger vessel was very close to Keller-Miksis equation's results (Fig. 3). Although, the overall bubble oscillation amplitude was reduced as it drifted towards the vessel wall (Fig. 5). However, the circumferential and shear stresses were increased almost linearly (Fig 6 & 7). Also, the bubble center moves due to asymmetric oscillation for bubble placed away from axis and close to vessel wall (Fig 4).

REFERENCES:

1. Hosseinkhah et. al., A three-dimensional model of an ultrasound contrast agent gas bubble and its mechanical effects on microvessels , Phys Med Biol, 57, 785-808 (2012)