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A Simulation of Vessel–Clamp Interaction: Transient Closure Dynamics

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Abstract

Cross-clamping of aorta is routinely performed in cardiac surgery. The objective of this study was to simulate cross-clamping of the aorta to elucidate the perturbation of stresses in the wall (solid mechanics) and lumen of the vessel (fluid mechanics). Models of the aorta and clamp were created in Computer Assisted Design and Finite Element Analysis packages. The vessel wall was considered as a non-linear anisotropic material while the fluid was simulated as Newtonian with pulsatile flow. The clamp was applied to produce total occlusion in approximately 1 s. A cylindrical and rectangular geometry for the clamp were considered. High jet speed and flow reversal were demonstrated during clamping. It was found that the clamp design and vessel wall anisotropy affected both the fluid wall shear stress (WSS) and solid stresses in vessel wall. The maximum wall stresses increased by about 170 and 220% during closure in the cases of plate and cylindrical clamps, respectively. The plate clamp design was superior for reduction of both solid stresses as well as fluid shear stresses. The cylindrical clamp causes much larger stresses than the plate clamp in each of the stress components; e.g., radial compression of -180 vs. -50 kPa. Vibrations, flow and WSS oscillations were detected immediately before total vessel occlusion. The present findings provide valuable insights into the mode of tissue injury during clamping and may also be useful for improving surgical clamp designs.

Keywords

Fluid; solid interaction; Endothelium; Intramural wall stress; Wall shear stress; Surgical clamps

INTRODUCTION

Vascular cross-clamping is necessary in many cardiovascular surgeries such as coronary bypass, aorta repair, and valve procedures. The purpose is generally to arrest blood flow to

prevent bleeding. Although the current clamps are marketed as so called “atraumatic”, evidence suggests otherwise. Experimental studies have found that clamping of various degrees (i.e. from notch 3 to 6 setting on the clamps) causes various degrees of damage to the artery, from endothelium denudation/delamination to severe intima injury as observed under light and scanning electron microscope.^{1,6,17–20} Since aortic clamping alone is performed in about half million patients per year in the U.S., a safer clamp would affect many patients.

The highly transient fluid dynamics, distortions of the vessel wall and the complex mechanics of clamping are important variables in the occlusion process. The vessel deformation and stress analysis can provide valuable insights into the stimuli of tissue injury.^{3,10,17,18} Realistic computational models may lead to significant improvement of vascular clamp design, and are thus of potential interest for surgeons and clamp manufacturers. Despite the apparent clinical significance, however, there have been few computational studies of clamping. To our knowledge, the only clamp simulations have been provided by Gasser *et al.*¹⁰ and Calvo *et al.*,² both of which did not include fluid or fluid–solid interaction (FSI). Furthermore, both models did not reach complete occlusion (92% in Gasser *et al.* and about 60% in Calvo *et al.*) and cited difficulties in achieving convergence when the vessel was highly deformed. Arterial clamping involves large deformations, contact and FSI which are computationally demanding.

The objective of the present study was to develop a computational model that overcomes these limitations; i.e., clamp capable of full closure, incorporates FSI and is based on anisotropy of the vessel wall. Since total occlusion was achieved in the present study, the closure dynamics was analyzed. It was found that the clamp design and vessel wall anisotropy affect both the fluid wall shear stress (WSS) and solid stresses in the vessel wall. Vibrations and WSS oscillations were detected immediately before total vessel closure.

METHODS

Governing Equations

The computational models were created in a Computer Assisted Design package—Pro/Engineer and solved in a well validated Finite Element Analysis package—ADINA. The governing equations for the fluid domain are the Navier–Stokes & Continuity equations.⁷

$$\frac{\partial \vec{V}}{\partial t} + \vec{V} \cdot \vec{\nabla} \vec{V} + \frac{\vec{\nabla} P}{\rho} - 2 \frac{\eta}{\rho} \vec{\nabla} \cdot D = \vec{0} \quad (1)$$

$$\vec{\nabla} \cdot \vec{V} = 0 \quad (2)$$

where $\vec{V}$ is velocity, P is pressure, ρ is density, η is viscosity, D is the rate of deformation tensor, $\vec{\nabla}$ is the gradient operator.

The governing equations for the solid domain are the Momentum and Equilibrium equations⁸:

$$\rho a_i - \sigma_{i,j,j} - \rho f_i = 0 \quad \text{in} \quad {}^s\Omega(t) \quad (3)$$

$$\sigma_{ij}n_j - t_i = 0 \quad \text{on} \quad {}^s\Gamma(t) \quad (4)$$

where a_i is acceleration of a material point, f_i is force per unit mass, ${}^s\Omega(t)$ is the structural domain at time t , t_i is surface traction vector, σ_{ij} is stress.

Material Models

For the present application, the aorta was modeled as soft, non-linear material and the clamp was modeled as rigid. Isotropic and anisotropic material properties were considered for comparison. For the isotropic vessel material, the neo-Hookean non-linear model was used. The Strain Energy Function has the form:

$$\bar{\psi}_{\text{iso}}(\bar{I}_1) = \frac{C}{2}(\bar{I}_1 - 3) \quad (5)$$

where C is a material parameter. For the anisotropic material, the strain energy was modeled as the sum of two components: isotropic and anisotropic:

$$\bar{\psi}(\bar{C}, A_1, A_2) = \bar{\psi}_{\text{iso}}(\bar{I}_1) + \bar{\psi}_{\text{aniso}}(\bar{I}_4, \bar{I}_6) \quad (6)$$

This form obviously reduces to the isotropic case if only the first term is considered. The isotropic behavior was modeled using the neo-Hookean model described in Eq. (5).

The anisotropy was modeled by the following strain energy density¹¹:

$$W_o = \frac{K_1}{2K_2} \{ \exp[k_2(J_4 - 1)^2] + \exp[k_2(J_6 - 1)^2] - 2 \} \quad (7)$$

where $J_4 = I_4 I_3^{-1/3}$, $J_6 = I_6 I_3^{-1/3}$, $I_4 = C_{ij}(n_a)_i(n_a)_j$, $I_6 = C_{ij}(n_b)_i(n_b)_j$, C_{ij} is the Cauchy–Green deformation tensor. The material parameters were adapted from literature¹¹ as $k_1 = 2.86$, and $k_2 = 0.84$, $C = 200$ kPa.

The fluid was modeled as Newtonian, incompressible with pulsatile flow of 60 beats per minute. The density and viscosity were 1050 kg/m³ and 0.004 kg/m·s respectively.²² For the wall interface, we assume no slip between fluid and the wall and no permeability of the vessel wall. Two kinds of clamps were modeled: a rod-like clamp (cylindrical), similar to a clamp considered by Gasser *et al.* and a rectangular-shaped clamp. The stresses in the vessel wall as well as the fluid WSSs imposed by the two clamps were compared. The clamp was applied through a linear displacement time function to produce total occlusion in about 1 s.

Geometry and Mesh

The vessel segment was 80 mm long and 16.5 mm in diameter, representative of a typical ascending aorta in two dimensions.¹² The length was chosen to facilitate computation and to allow re-circulations to develop. The rectangular clamp was 2.5 mm wide while the cylindrical clamp was 2.5 mm in diameter for direct comparison of the two geometries (Fig. 1).

The vessel wall was discretized with 80 elements axially and 2 elements through the thickness. The clamp was designed with rounded edges to avoid stress concentrations caused

by sharp corners. It was discretized with 10 elements along the length and 4 elements along the thickness. The fluid was discretized with 160 elements axially and 20 elements radially. Because of symmetry, only half of vessel and clamp was simulated.

Contact Conditions

To model the interaction of boundary of clamp with the boundary of the artery, contact or multi-body interaction was setup between the approaching clamp and outer surfaces of the vessel. Penetrations between the clamp and vessel elements were not allowed by the contact algorithm. Once penetration was detected, the overlapping elements were returned to their positions at the previous time step. Contact was also setup on the opposing vessel walls as they came into contact during closure. Frictionless contact was assumed between the arterial and the clamp boundaries.

Fluid–Structure Interaction Simulations

Fluid–structure interfaces were defined at the lumen surfaces of the vessel, and the wall boundaries of the fluid. The ALE (Arbitrary Lagrange–Eulerian) method was used. This method allows the fluid mesh to deform around the moving vessel, which is necessary for this application with the large deformation. Instead of using either a single Lagrangian approach or a single Eulerian approach, the ALE describes the motion of fluid in a moving reference frame with the sole constraint that the velocity on the fluid–solid boundary must equal to that of the boundary. The velocity of the reference frame is usually neither the fluid particle velocity such as in a pure Lagrangian description nor zero in a pure Eulerian description.

The Navier–Stokes equations for the fluid and the momentum and equilibrium equations for the solid are coupled on the fluid–solid interface via the kinematic and dynamic conditions as follows:

$$\underline{\mathbf{d}}_f = \underline{\mathbf{d}}_s \quad (8)$$

$$\mathbf{n} \cdot \underline{\boldsymbol{\tau}}_f = \mathbf{n} \cdot \underline{\boldsymbol{\tau}}_s \quad (9)$$

where $\mathbf{n}$ denotes the normal vector of the interface. The symbols $\underline{\mathbf{d}}_f$ and $\underline{\mathbf{d}}_s$ denote the fluid and solid displacements. $\underline{\boldsymbol{\tau}}_f$ and $\underline{\boldsymbol{\tau}}_s$ denote the fluid and solid stresses. The underline indicates the variable is evaluated on the solid fluid interface. The solid and fluid models were coupled by the fluid nodal positions on the FSI interfaces. The displacements of the other fluid nodes were determined so as to preserve the initial mesh quality. The ALE modified governing equations for fluid flow were then solved. For the dynamic case, the fluid stresses were integrated along the fluid–solid interface and applied on the corresponding solid nodes.

The fully coupled two-way FSI model was solved. As shown in the flow chart (Fig. 2), the fluid model was solved at each time step. The wall mechanical state was then updated and the wall model was solved using fluid pressure and shear stress fields as input. At the next time step, with the newly updated vessel wall, we remesh the fluid domain, calculate nodal velocities, and solve the fluid model again. The solution process continued until solutions for solid and fluid nodes converge on the FSI interface and steady solution was achieved. Most of the computation time was taken during the final closure phase (15% of the displacement time function). The CPU time spent in the solid–fluid coupling was between 10–20 times more than the computing required for fluid only solutions (CFD only).

RESULTS

The flow field near closure is shown in Fig. 3 where strong FSI is observed. A dramatic reduction of flow velocity near closure is observed. Figure 3a shows the overall flow field. Figures 3b and 3c reveal important flow features such as the flow reversal upstream and recirculations downstream. WSSs induced by the two types of clamps are shown in Fig. 4 where the WSS spiked before closing. The plot indicates a high WSS in the occlusion region as expected.

The WSS in isotropic and anisotropic vessel wall is shown in Fig. 5. The WSS near occlusion at clamp center in the anisotropic vessel had two large peaks. Flow oscillations were also observed during closure which causes significant oscillatory shear stress. The WSS around the edge of the clamp had a single peak near closing, followed by oscillations.

The maximum flow velocity during clamping is shown in Fig. 6. A small initial disturbance was observed as the clamp came in contact with the vessel wall and prompts a wave through the flow field. Figure 6b shows the flow velocity upstream of the clamp. The velocities immediately upstream (2 mm— similar to the width of clamp, from occlusion) and further upstream (15 mm from occlusion) were plotted. During the final stage of clamping (after about 90% occlusion), flow reversal in the upstream section was observed as indicated by the negative flow. The predicted maximum flow velocity of 1.2 m/s is consistent with the aortic flow experimental data in the range of 0.8–1.4 m/s.²¹

The solid stresses in the vessel wall induced by the two types of clamps were compared. The maximum Von Mises stresses with displacement driven clamp is plotted in Fig. 7. It is shown that the maximum Von Mises stresses in the vessel wall (lumen side of clamped area) are much higher with the cylindrical clamp. The stresses spiked as the tissue was being squeezed during closure (Figs. 7 and 8). The cylindrical clamp causes much larger stresses than the plate clamp in each of the stress components; e.g. radial compression of –180 vs. –50 kPa (Figs. 8a and 8b). For the cylindrical clamp, the data show that the radial compression and axial stresses were large at closure: –180 kPa and 170 kPa, respectively. In comparison, shear was modest at about 20 kPa. For the plate like clamp, the simulations revealed that axial stresses (90 kPa) were larger than compressive –50 kPa and shear stresses (20 kPa). Figure 8c shows the shear stress caused by the plate and cylindrical clamps.

Figures 9a and 9b show the maximum stresses at the inner and outer vessel wall caused by cylindrical and plate like clamps. It is seen that the maximum stresses in the vessel wall (lumen side of clamped area) are much higher with both clamps. The maximum Von Mises stress spiked as the tissue was being squeezed near closure.

Figure 10 shows the isotropic and anisotropic vessel wall velocity during clamping. The wall velocity changed rapidly towards closure, just before the wall motion ceased. The anisotropic vessel had higher rate of change of wall velocity or acceleration. This large fluctuation of velocity indicates vibration of the vessel wall.

DISCUSSION

This is the first model of the arterial cross-clamping process with FSI capable of full closure that takes into account the anisotropic properties of vessel wall. Furthermore, these simulations combined FSI with multi-body interaction between rigid surgical instruments and anisotropic soft tissue. Solids and fluids have distinctively different types of governing (momentum) equations. There are numerous technical challenges, including large

deformations, non-linear anisotropy and FSI. The convergence and equilibrium on the solid–fluid interface are also extremely computationally demanding.

The simulations show that the clamp geometry and vessel wall anisotropy affect fluid WSS and solid stresses in vessel wall. Vibrations and WSS oscillations were detected immediately before total vessel closure.

Closure Dynamics

Figures 3 and 4 illustrate the dynamic and highly transient nature of the fluid during the closure process and support the importance of implementing the FSI model. The fluid shear plot indicates that high shear stresses exist in the occlusion region as expected (Fig. 4). Also, fluid shear stresses became higher towards vessel wall where endothelium resides. It is well known that abnormally low or high WSS have effects on endothelial biology.^{4,5,8,9,15,16,22} It can be seen from Fig. 4 that the high shear stress regions extend further downstream in the flow field. WSS spiked during closure in the cylindrical clamp case but not in the plate clamp. (Figs. 5 and 6).

The maximum stresses in the arterial wall were higher on the inner than the outer surface at full occlusion (Figs. 7 and 9). This is important as the endothelium resides on the lumen side of the artery. It was found that the cylindrical clamp caused much larger stresses than the plate clamp in each of the stress components: e.g. radial compression of -180 vs. -50 kPa (Figs. 7 and 8). In the case of the cylindrical clamp, the simulations show that the radial compression and axial stresses were of largest magnitude (-180 and 170 kPa, respectively). Shear stress was relatively small, maximum of 20 kPa. Therefore, the injury mechanism is likely to stem from compression and axial stresses caused by the clamps (Fig. 8). The vessel may be more prone to injury by the large compressive stress during closure since the aorta normally does not experience significant compression. In the case of the plate like clamp, the simulations show that axial stresses were larger than compression and shear stresses (90 kPa vs. -50 kPa and 20 kPa, respectively) (Fig. 8). For the purpose of reducing both fluid shear stresses and solid stresses (Figs. 4, 7 and 8), the plate clamp design was superior. The most likely reason is that under the rectangular clamp, there is an elongated contact area which reduces and smoothes the stress. One of the objectives of this study was to demonstrate the capabilities of the models to study the effects of certain design features. This may serve as a guide during the design phase of medical devices. Future models could be developed as optimization tools. These simulations may save both time and cost of building and testing various prototypes.

Although this model was used to study the cross-clamping process, the FSI model can also simulate various stages of stenosis, 80, 90, 95%, etc. (Figs. 3 and 4). These may yield insights into the fluid mechanics of various degrees of arterial stenosis. Figure 3a demonstrates that at about 80% occlusion, flow reversal in the upstream section was observed as indicated by the negative flow. Severe stenosis is highly conducive for flow reversal. This has important implications as reverse flow can cause endothelium to produce more superoxides which can scavenge nitric oxide.^{13,14}

Critique of the Model

The plate like clamp in the current model has a smooth contact surface. Actually surgical clamps have teeth like surfaces. A model incorporating this feature is under development. Future models will also include contact with friction. Furthermore, the current model is two dimensional and the flow field did not include the radial component. Under these conditions, it is expected that the axial flow velocity is order of magnitude greater than circumferential and radial components. As the Calvo *et al.*² and Gasser *et al.*¹⁰ studies demonstrated that

even solid models alone without fluid coupling were extremely difficult to reach closure in 3-D. The FSI coupling would make this problem more difficult with the fluid mesh in-between the opposing solid walls. Since the main focus of this study is on closure dynamics of fluid WSS and solid stresses, this model focuses on 2-D with FSI coupling. The FSI simulations are 10–20 times more computationally expansive than CFD or fluid only simulations. Future studies will consider a 3-D formulation with FSI.

The clamping was initiated at the onset of systole. Surgeons do not usually synchronize clamping with the cardiac cycle. It is possible that the timing may affect the results somewhat. Slowed clamping during the final phase could possibly reduce the temporal gradient of WSS and intramural stresses in both clamp cases. This should be investigated in subsequent studies. The current model did not incorporate plaque structures in the vessel wall which is common in patients requiring cross-clamping.^{18,20} Future models should include the non-homogenous structure of the vessel wall. Finally, the cylindrical clamp geometry is not used in practice. It was included for comparison with the Gasser *et al.* and Calvo *et al.* studies.

Comparisons with Previous Studies

In a clamping study by Gasser *et al.*,¹⁰ the model did not include fluid or FSI and did not attain total occlusion. In another clamping model published by Calvo *et al.*,² the model also was not coupled with fluid and achieved occlusion level of about 60%. The authors reported that axial stresses at the inner wall were significantly increased and were cited as the potential culprit for vessel injury. During the closure process, Fig. 7 showed that the overall effective stresses rose to nearly three times as high as the magnitude before closure. Importantly, Gasser *et al.* and Calvo *et al.* studies did not identify the high compressive stress as their models did not reach closure during which the compressive stress spikes. As shown in Fig. 8b, the compressive stress was nearly negligible until closure was reached. It was only during the closure that the compressive stress rose dramatically. Therefore, the current study revealed a critical mode of tissue injury that was missed by previous models. The vessel may be more prone to injury by the spike of compressive stress during closure since the aorta normally does not experience significant compression.

Additionally, the current study found very interesting fluid phenomena during the closure process; i.e., the flow reversal and spike of WSS, etc. The resulting high temporal gradient of WSS may activate platelets and initiate thrombus formation. These were not reported in Gasser *et al.* and Calvo *et al.* studies, as their models did not include fluid. In addition, the current study considered different clamp designs which was not evaluated in the previous studies.

Significance of the Study

The present model simulates an aortic cross-clamping process with emphasis on closure dynamics of both solid and fluid. Both the fluid and solid mechanics during the closure process are found to be highly transient and dynamic. The utility of the model is to enable optimization of the design of clamps that may cause less trauma to arteries. Clearly, the plate clamp design is superior for reduction of both solid stresses and fluid shear stresses. The vibrations during the closure process may be used for development of a clamp with a sensor that can detect such vibrations to avoid over-closure and hence injury to the vessel wall. Furthermore, this model may serve as a basis for virtual clamping where surgeons can examine various surgical scenarios *in silico* and predict the mechanical stresses in the vessel wall.

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FIGURE 1.

The geometry and mesh of clamp, vessel and internal fluid at various stages of clamping: (a) earlier stage and (b) later stage of clamping.

FIGURE 2.

The fluid–structure interaction algorithm flow chart. P is pressure, V is velocity, t is time, dt denotes a time step.

FIGURE 3. The flow field and detailed flow features: (a) overall flow field, (b) flow reversal profile, and (c) flow re-circulation.

FIGURE 4.

(a) Fluid wall shear stresses induced by the plate like clamp with zoom in view and (b) comparison of wall shear stresses data induced by the two kinds of clamps.

FIGURE 5. Wall shear stresses on isotropic and anisotropic vessel endothelium by plate like clamp: (a) at center of clamped area and (b) at edge of clamped area.

FIGURE 6.

(a) The maximum fluid flow velocity during clamping by plate like clamp at occlusion area and (b) the maximum fluid flow velocity during clamping at near occlusion (2 mm away) and upstream (15 mm away) areas. Arrow denotes closure.

FIGURE 7.

The vessel wall stresses induced by the two kinds of clamps: (a) stress profile in vessel wall during closure with plate like clamp, (b) stress profile in vessel wall during closure with cylindrical clamp, and (c) the maximum stress in vessel wall induced by the clamps.

FIGURE 8.

The axial, radial and shear stress components caused by the plate and cylindrical clamps. Closure occurs at 1 s.

FIGURE 9.

(a) The maximum stresses at the inner and outer vessel wall caused by cylindrical clamp and
 (b) the maximum stresses at the inner and outer vessel wall caused by plate like clamp.

FIGURE 10.

Vessel wall velocity of isotropic and anisotropic vessel during clamping by plate like clamp. A rapid change of wall velocity indicates vibration just before closure.