Shock Wave Induced Damage in Soft Tissue

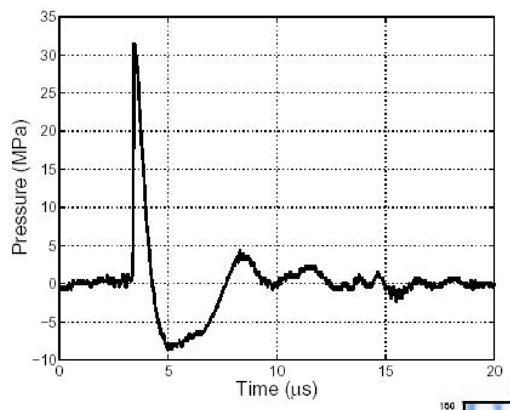
K. Weinberg, M. Ortiz and T. Colonius

Technical University of Berlin

California Institute of Technology

Problem: Shock-Wave Lithotripsy (SWL)

In SWL, a very common, non-invasive method to destroy kidney stones, the kidney is blasted with high intensity sound waves generated outside the patient and focused on the stone.

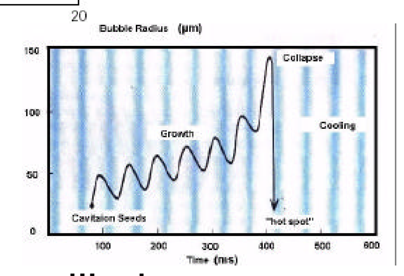


focused (direct) shock waves:

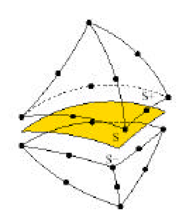
- compression induced cracks
- stone spallation by reflection
- tissue shearing
- tension "tail" of the pressure impul

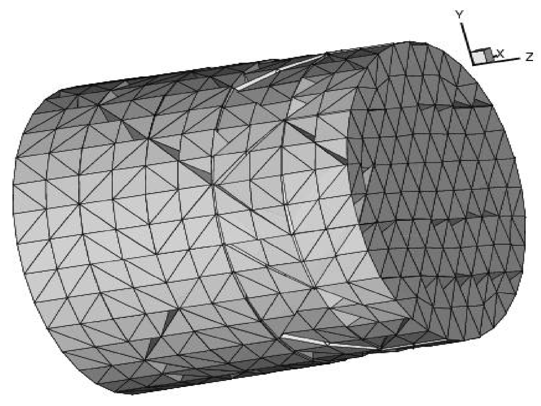
indirect shock waves:

- cavitation - bubble expansion and collapse
- pitting by "jets" of collapsing bubbles
- tension waves by reflection – dilatation of vessels and capillaries



Our numerical simulations of stone spallation use a fragmentation technique of [Pandolfi & Ortiz, 2002] with special cohesive elements.





Journal of Urology (1992):
"SWL is a form of renal trauma"
Shock waves focus at stone but damage tissue!

Material Model: Kidney Tissue

The elastic behavior of the soft tissue is constrained by a deviatoric stress measure corresponding to a shear stress.

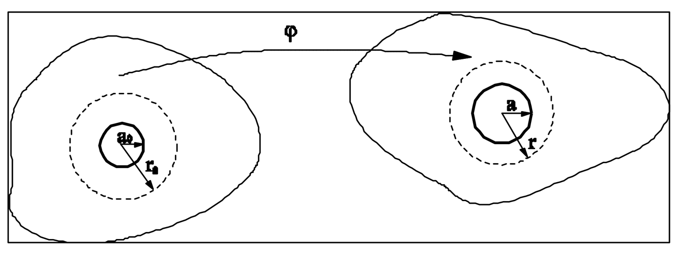
logarithmic plastic strain: e^p

The volumetric expansion of the body $V = JV_0$ with $J = J^e J^p = \det(\mathbf{F})$ is induced by hydrostatic tension.

logarithmic plastic dilatation: J^p

The tissue material is incompressible, irreversible volumetric expansion is only induced by cavitation of (empty) bubbles.

$$\frac{d}{dt}\frac{4p}{3}(r^3 - a^3) = 0 \quad \rightarrow \quad \dot{e}^p = \left|\frac{\partial \dot{r}}{\partial r}\right|$$



Formulate the corresponding energy of dissipated volumetric work and the kinetic energy of expanding bubbles as a function of radius, expansion velocity and cavity size distribution.

The plastic dilation relates to micro-mechanic via the volume of the cavities, e.g., for N cavities with radius a:

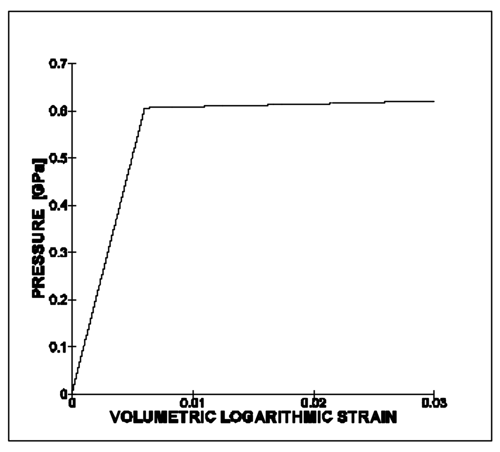
$$J^p = \log\left(1 + \tfrac{4}{3} p\, N\left(a^3 - a_0^3\right)\right)$$

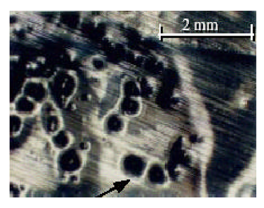
Use an incremental solution procedure and define in every time interval an update-energy function in terms of logarithmic plastic variables.

$$f_n(\mathbf{F}_{n+1}, e^p_{n+1}, \mathbf{M}, q^p_{n+1}) = W^e(\mathbf{F}^e_{n+1}, J^p_{n+1}) + W^p(e^p_{n+1}, J^p_{n+1}) + K\left(\dot{j}^p, J^p_{n+1}\right) + \Delta t y^*\left(\Delta J^p, \Delta e^p, \Delta t\right)$$

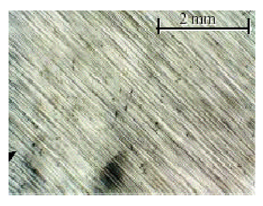
Compute the updated internal variables by: $W_n = \min_{e^p_{n+1}, \mathbf{M}, J^p_{n+1}} f_n$

In tension the hydrostatic pressure s bounded by a critical cavitation pressure $p = p(E, v, s_y, t, \ldots)$, in compression the material is pressure insensitive.

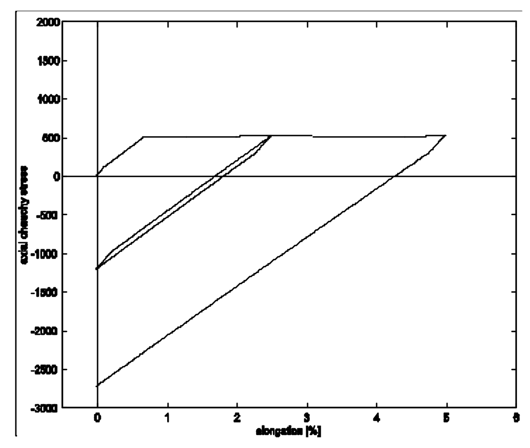




[Lokhandwalla, 2001]
pitting by cavitation



no effect of overpressure



The model accounts for the effects of micro-inertia, rate-sensitivity and viscosity.

Numerical Simulation

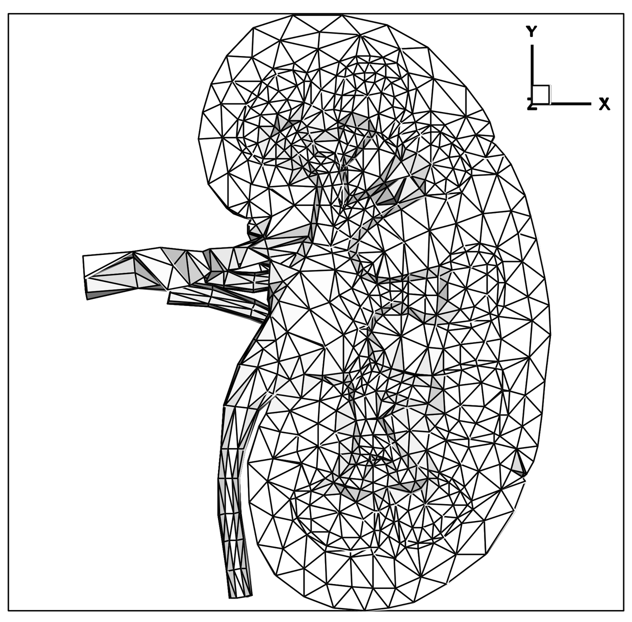
- Elastic material data (elastic modulus 0.05 … 0.3 MPa, viscosity 0.001 … 0.007 Pa s) are known, but only few experiments determine inelastic properties.
- Different parameters for different regions of the kidney are not available, all data are homogenized for "the kidney".
- Parametric studies on the effect of varying material data (uniaxial strain):

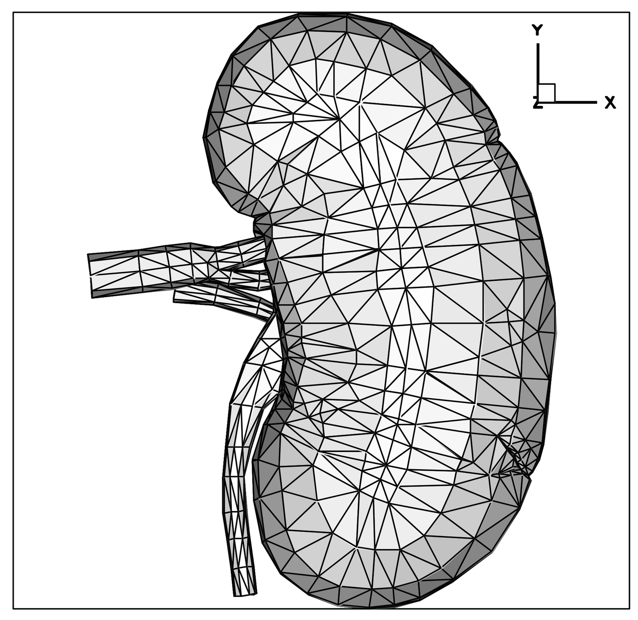
body tissue kidney tissue fat

propagation of (damage inducing) stress waves

- First finite element mesh bases on geometrical data purchased from a company.

mid-surface and full model



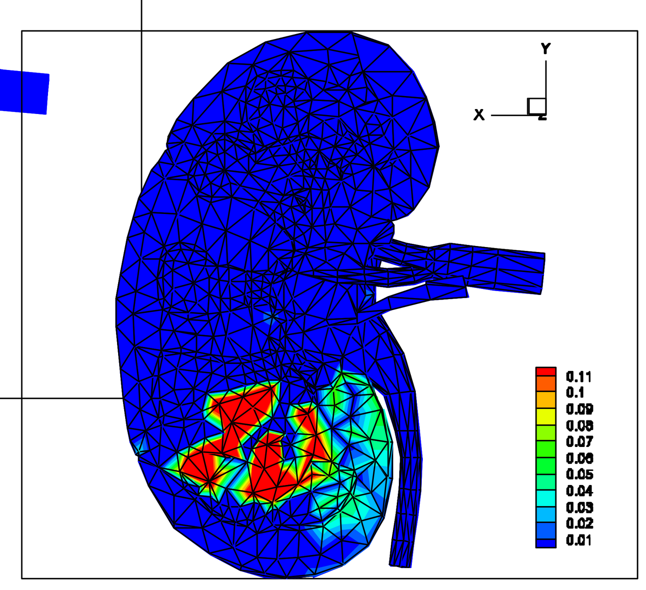


- functional tissue (renal cortex, medulla, …) : new material model
- non-functional tissue (ureter, main arteries and veins): non-linear elastic material

Fig 2

mid-surface

outer surface



- damage in a pig kidney experiments by [Cleveland et al., 2000]
- color: irreversible volumetric expansion (damage)