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Abstract

In this study, we develop some steps of a ray tracing method for shock wave modeling in a lithotripter. A realistic 3D CAD human model is used for this purpose. Every individual ray undergoes refraction when crossing a triangulated boundary between two adjacent tissues with different impedances. The transmission angle is uniquely defined for 2-manifold tissue objects.

Within every tissue, a nonlinear distortion of a shock pulse is taken into account using a simplified quasi-planar approach. The viscous damping is neglected due to the lack of material data. The ray-tube model is not used. Instead, we apply the Voronoi partition in every transversal plane in order to find the wave amplitude and intensity at a particular point. Some preliminary simulation results are reported related to a shift of a focal point in the focal plane and the effect of body size on the focal pressure. Major advantages of our method are the high speed of computations and thus a potential ability to use a patient-specific CAD model for calibration purposes in real time.

Development of a 3D model in COMSOL was also used for this study to investigate how Gaussian pulses generated from a point source would propagate through human tissue. However, the results were insignificant as the software seemed unfit to be used with high frequency pulses.

Review
Shock wave lithotripsy has been used since 1980 as a noninvasive medical procedure to help patients pass kidney stones, which are a formation of stones within the urinary tract [1]. Shock waves are high energy pressure pulses generated by a quick energy release [1].

Finite element analysis and simulation software can be used to model the shock wave lithotripter and the pulse that is generated from the spark. We began an investigation into the use of COMSOL as a means to theoretically measure the pressure at certain points in the model.

Ray tracing can be used as an approximation to solving the full wave equation with Green’s function or the Westervelt equation [2]. Ray tracing can be used as an ultrasound simulator, but here we will use ray tracing to follow the rays and attempt to determine the accuracy of lithotripters computationally.

Introduction

Kidney stones are created when crystal forming substances – like calcium, oxalate, and uric acid – reach a threshold concentration that can’t be diluted. This leads to a crystal substance within the urinary tract [3]. Depending on the size of the kidney stone, they may not be able to travel through the urinary tract unimpeded.

A device called a lithotripter can be used to try to break up the kidney stone when the kidney stone cannot be passed on its own. A lithotripter produces acoustic waves produced from a shock wave. However, over time, the lithotripters have been developed to produce high amplitude and tightly focused shock waves that cause damage not only to the kidney stone, but to the peripheral tissues [3]. The method of focusing the lithotripter depends on CT scans that determine where the kidney stone is. When the shock waves are produced, the doctors give the
patient a CT scan to determine whether the kidney stone fragmented. If it didn’t, they just run the lithotripter again, and repeat this process until the stone has been fragmented or destroyed.

There is a question of how accurate lithotripters are. Since the lithotripters may have to be ran multiple times to fragment a single stone, there is some belief that they aren’t focusing at the intended spot. The focus of this research is to build a computational model of the lithotripter and the human body to see if the focal point matches the intended focal point.

Justification for ray tracing

Ray tracing does not solve the full wave equation, but rather uses a simplified geometrical procedure, which neglects the diffraction effects of the acoustic field entirely. However, the wave refraction is still in place. Ray tracing performs reasonably well when the wavelength (for continuous radiation) or pulse length (for pulses) is much less than a typical geometry scale. In lithotripsy, the pulse length is on the order of 1-2 mm [4]. This value considerably exceeds geometrical tissue variations observed along the focusing beam path. Thus, ray tracing is potentially applicable to the present problem.

Ray tracing for 3D CAD tissue objects

In its simplest form, a 3D CAD tissue object is characterized by an array of nodes $P$ and an array of triangles (connectivity), $t$, which define its surface. The human model we use has an “onion” topology. Namely, all tissue objects are closed 2-manifold shells, which neither intersect nor touch each other, but may contain other tissue objects. For example, the outermost “skin” object contains all other tissue objects. The kidney objects are located inside the “average body” shell; which surrounds all internal organs and fills space between them with “average body”
properties. This onion topology allows us to uniquely define outer normal vectors for every 3D tissue object and perform ray tracing in a correct and effective way.

Consider a ray with a unit direction vector $\mathbf{i}$ incident upon a triangular facet of a tissue object with a unit outer normal vector $\mathbf{n}$ from tissue outside as shown in Fig. 1. This particular facet is identified using a vectorized ray-triangle intersection algorithm [5] and checking all triangles of the object or a group of selected triangles only. With reference to Fig. 1, one has for a local orthogonal (but non-normalized) basis $\mathbf{k}$, $\mathbf{l}$, $\mathbf{m}$ at the incidence point, angle $\alpha$, and non-normalized directional vectors $\mathbf{r}$ and $\mathbf{t}$ for reflected and transmitted rays, respectively,

The first investigation into a solution to this problem was through a software called COMSOL Multiphysics, which is a finite element analysis and simulation solver. Our initial plan was to import a model of the human body into the software, run acoustic simulations on the model, and interpret the results.

Our next step was to develop our own solution through programming in MATLAB. This involved inputting the data for the human model into MATLAB, creating a simulation of a lithotripter, and running simulations with waves to see how they moved through the human body and graphing the results.

Methods

The lithotripter was modeled in COMSOL with a point source that serves as the source for the Gaussian pulse that ranges from 10KHz to 1MHz. The body was modeled using files created in MATLAB as .mat files that were converted to .stl files.
The lithotripter will be modeled by a parabolic reflector in the programming solution, since the lithotripter is a parabolic reflector within a water bath with a spark plug that generates the shock wave. The spark plug is located in the middle of the parabolic reflector, but since it’s only purpose is to generate waves that bounce back from the reflector, for this research we will ignore that. Instead, we will create waves that begin from the parabolic reflector, assuming that the spark plug has already generated the waves.

Figure 1. Example of reflector

The human body will be represented by a triangular mesh that is primarily composed of layers of fat, muscle, another fat layer, and the kidney.
The reason why the mesh is constructed this way is because the other tissues aren’t really important for this machine. By placing the lithotripter in the back, you can avoid the major bones from the rib cage, and the organs in the front of the body. Also, we are generally only interested in the focal point at the medulla of the kidney, and aren’t concerned with the tissues after the rays pass the kidney, so there isn’t a need to model the peripheral tissues besides those already considered.

The lithotripter is modeled as a portion of a sphere with the center of the sphere as the focal point. This assures that the distance to the focal point will be the same for all rays. The rays are modeled as simple vectors with an origin and direction. The origin is defined as the vertices that make up the triangular mesh of the parabolic reflector. The direction is just the difference between the origin and the focal point. At each interface between tissues, where the initial fluid
is assumed to be water since the lithotripter is in a water bath, the origin for the ray becomes the point where it intersects the interface. The new direction is determined by the angle of the ray incident to the interface and the two acoustic properties of the two tissues that make the interface.

Ray Transmission

Overview: When a ray hits a plane, which is an example of a ray hitting the boundary of two different fluids with different acoustic properties, part of the ray will reflect off the plane and part of the ray will transmit through the plane. The angle of the reflection ray on a plane will be the same as the angle of the incident ray on a plane. However, the transmitted ray angle with respect to the incident ray will depend on the acoustic properties of the fluids on both sides of the boundary [4].

\[ l = \mathbf{m} \times \mathbf{i}, \quad k = \mathbf{l} \times \mathbf{m}, \quad r = (\mathbf{i} \cdot \mathbf{k}) \mathbf{k} - (\mathbf{i} \cdot \mathbf{m}) \mathbf{m}, \quad \cos \alpha = (\mathbf{i} \cdot \mathbf{k}), \quad t = \cos \beta \mathbf{k} + \sin \beta \mathbf{m} \]

Figure 3. Example of a ray hitting a plane
The angle $\Theta_i$ is the angle of the incident ray on a plane. The angle $\Theta_r$ is the angle of the reflected ray on a plane. The angle $\Theta_t$ is the angle of the transmitted ray on a plane. The variable $c_1$ is the phase speed of fluid one, and $c_2$ is the phase speed of fluid two. The variable $\rho_1$ is the equilibrium density of fluid one, and $\rho_2$ is the equilibrium density of fluid two.

Acoustic impedance of fluid one, $r_1$, is the product of $\rho_1 * c_1$, and the acoustic impedance of fluid two, $r_2$, is the product of $\rho_2 * c_2$. The acoustic impedance will be valuable when we calculate the angle of complete transmission and the power transmission coefficient [4].

**Reflection:**

The reflected angle, $\Theta_r$, will equal the value of the incident angle, $\Theta_i$.

$$\theta_r = \theta_i$$

This makes sense, because the acoustic properties of the fluid will remain the same if the ray is reflected, so the reflected ray should behave similar to the incident ray in a fluid of the same property. The amount of the intensity conserved from the incident ray to the reflected ray is given by the Rayleigh reflection coefficient, $R$.

$$\left( \frac{r_2}{r_1} \right) \frac{\cos(\theta_r)}{\cos(\theta_i)} = R$$

$$\left( \frac{r_2}{r_1} \right) + \frac{\cos(\theta_r)}{\cos(\theta_i)} = R$$

When $R = 1$, there will be complete reflection. When $R = 0$, there will be complete transmission. Note that $R$ is based on the transmission angle as well as the reflection angle, so the transmission angle may have to be calculated before you can calculate $R$ [4].

**Transmission:**

The transmitted angle, $\Theta_t$, varies on the parameters of the fluids involved.

$$\frac{\sin(\theta_r)}{c_1} = \frac{\sin(\theta_t)}{c_2}, \quad \sin \theta_r = \left( \frac{c_2}{c_1} \right) \times \sin(\theta_i)$$
By applying the identity, $1 = \cos^2(\theta) + \sin^2(\theta)$, we can solve the equation above for $\cos(\theta)$ and apply this to our transmission angles to get:

$$\cos(\theta_t) = \sqrt{1 - \left(\frac{c_2}{c_1}\right)^2 \sin^2(\theta_i)}$$

This equation is valuable because it gives an idea of conditions where the angle will be real or when the angle will be imaginary. This brings the idea of a critical angle, where the critical angle determines whether the transmission will be real or imaginary depending on the fluid properties. This is because if $c_1 < c_2$, the incident angle may be $> 1$, and the root will be negative and have an imaginary result [4].

$$\sin(\theta_c) = \frac{c_1}{c_2}$$

**Transmitted Angles:**

**Case 1: $C_1 > C_2$**
- The value of the sine will never increase over one, and the product is guaranteed to be less than 1, so $\Theta_t$ is real and less than $\Theta_i$.
- The transmitted angle is bent toward the normal to the boundary.

**Case 2: $C_1 < C_2$, and $\Theta_i < \Theta_c$**
- The transmitted angle is bent away from the normal to the boundary.

**Case 3: $C_1 < C_2$, and $\Theta_i > \Theta_c$**
- This is the case with an imaginary result, so there is complete reflection, $R = 1$.

The relationship between the incident ray, the transmitted ray, and the reflected ray is observed by the conservation of energy equation where:

$$1 = R_\pi + T_\pi$$

Where $1$ is the total proportion of the incident ray, or 100% of the intensity. Where $R_\pi = |R|^2$, or the square of the Rayleigh coefficient, $R$. The conservation of energy equation above still
supports the statement made earlier that when $R = 1$, there is complete reflection, and when $R = 0$, there is complete transmission. We can use the conservation of energy equation to solve for $T_\pi$, since we already know the equation for $R_\pi$ and $R$. We get the power transmission coefficient, $T_\pi$, as a result [4].

$$T_\pi = \left\{ \frac{r_2}{r_1} \right\} \cos(\theta_i) \cos(\theta_f) \left( \frac{r_2}{r_1} + \cos(\theta_i) \right)^2$$

Since there is complete transmission when $R_\pi$ or $R = 1$, we can calculate the formula for when there will be complete transmission, also known as the angle of intromission.

$$\sin(\theta_i) = \sqrt{\frac{1 - \left( \frac{r_1}{r_2} \right)^2}{1 - \left( \frac{\rho_1}{\rho_2} \right)^2}}$$

This angle will only exist if $r_1$ is less than $r_2$ and $\rho_1$ is less than $\rho_2$, or $r_1$ is greater than $r_2$ and $\rho_1$ is greater than $\rho_2$.

After the transmission angle has been calculated, the length of the ray can be calculated by normalizing the direction of the vector to a unit vector and multiplying it by the distance to the next interface [4].

Computational Human Model

A significantly simplified computational setup has been extracted from an accurate full-body CAD model VHP-Female version 2.1 intended for medical use [6-9]. This model includes 25 individual tissues and over 200 separate tissue parts, all extracted from the open-source Visible Human Project-Female© cryosection dataset of the National Library of Medicine and in the form of 3D CAD objects (2-manifold triangular surface meshes). A ~60 year old female subject has a height
$h$ of 162 cm measured from top of the scalp to the average center of both heels. The body mass $M$ computed using standard tissue densities [10] and assigning the average body shell, which includes internal tissues, the muscle density is 87.8 kg. The computed BMI is 33.5 (moderately obese). The model has a heart pathology.

The present simplified setup shown in Fig. 2a includes a low-resolution outer fat layer (yellow), a very thin muscle layer (individual muscle objects are not used), an average-body container (green, assigned here as fat; but it can be assigned muscle properties too), and the left kidney. Acoustic properties including the propagation speed, density, and a non-linear parameter $B/A$ have been acquired from Refs. [12, 13]. They are given in Table 1.

Nonlinear theory

The acoustic properties necessary for computation are speed of sound, and acoustic impedance for water, kidney, muscle, and fat. These measurements are:

Table 1: Values for Speed of Sound and Acoustic Impedance for different tissues [12]

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Speed of Sound [m/sec]</th>
<th>Acoustic Impedance [kg/sec*m^2] * 10^6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1480</td>
<td>1.48</td>
</tr>
<tr>
<td>Fat</td>
<td>1450</td>
<td>1.38</td>
</tr>
<tr>
<td>Kidney</td>
<td>1570</td>
<td>1.65</td>
</tr>
<tr>
<td>Muscle</td>
<td>1580</td>
<td>1.68</td>
</tr>
</tbody>
</table>

**Simplified nonlinear distortion model used in the paper**

The derivation of the nonlinear pulse correction (amplitude decay) will be given with reference to Fig. 4 below [12]. On the left of Fig.4, we show the original pulse (grey and green) and a non-linearly distorted pulse (green and red). Only the green path is physical and will remain. We need to find the corresponding amplitude decay.
Fig. 4. Accumulation of nonlinear effects over a path through a single tissue.

With reference to the blue and green triangles, we can see that they share a vertex, and have similar angles. This means that the two triangles are proportional, from this, we can set a ratio of the similar sides [6].

\[
\frac{x}{y} = \frac{\Delta}{l} \rightarrow xl = \Delta y
\]  
(1)

\[
xl = (P_0 - x)\Delta \rightarrow xl = P_0\Delta - \Delta x
\]  
(2)

\[
xl + \Delta x = P_0\Delta \rightarrow x(l + \Delta) = P_0\Delta
\]  
(3)

\[
x = \frac{P_0\Delta}{l + \Delta}, \quad y = P_0 - x
\]  
(4)

\[
y = \frac{P_0l + P_0\Delta - P_0\Delta}{l + \Delta} \rightarrow y = \frac{P_0l}{l + \Delta}
\]  
(5)

Next, we would like to find the ratio of the final pressure amplitude \( y \) to the initial pressure amplitude \( P_0 \).
\[
\frac{y}{P_0} = \frac{P_0 l}{l + \Delta} \rightarrow \frac{y}{P_0} = \frac{l}{l + \Delta} \rightarrow \frac{y}{P_0} = \frac{1}{1 + \frac{\Delta}{l}} \tag{6}
\]

Now that we have used geometry to find the relation between the initial pressure and \(y\), we need to apply the physics that shows how the wave degrades over time.

\[
t = \frac{\Delta x}{c_0} \tag{7}
\]

\[
c = c_0 + \Delta c \tag{8}
\]

\[
\Delta c = \left( \frac{B}{2A} \right) \left( \frac{P_0}{\rho_0 c_0} \right) \left( \frac{2A}{B} + 1 \right) \tag{9}
\]

\[
\Delta = t \Delta c \tag{10}
\]

\[
\Delta = \left( \frac{\Delta x}{c_0} \right) \left( \frac{B}{2A} \right) \left( \frac{P_0}{\rho_0 c_0} \right) \left( \frac{2A}{B} + 1 \right) \tag{11}
\]

\[
\Delta = \frac{\Delta x P_0}{c_0^2 \rho_0} \left[ 1 + \frac{B}{2A} \right] \rightarrow \Delta = \frac{\Delta x P_0}{2c_0^2 \rho_0} \left[ 2 + \frac{B}{A} \right] \tag{12}
\]

Now apply the derived \(\Delta\) to the previously derived pressure ratio.

\[
\frac{y}{P_0} = \frac{1}{1 + \frac{\Delta}{l}} \rightarrow \frac{y}{P_0} = \frac{1}{1 + \frac{\Delta x P_0}{2c_0^2 \rho_0} \left[ 2 + \frac{B}{A} \right]} \tag{13}
\]

Algorithm

Fig 5 shows this simplified simulation setup. In this study we did not model a complete parabolic reflector with a focal excitation. Instead, we considered an idealized spherical beam already formed by this reflector. Such a beam should be formed in the water bath near the body.

It is presumably converging at a point, which is the sphere center. To initialize the rays, we used an artificial spherical cap triangulated into nearly equal facets with an area \(A_0\), each as show in Fig. 2a. The ray intensity is proportional to the area of the facet. Every ray is emanating from
the facet’s center in the direction of the normal vector, toward the focal point. The number of rays varies from 200 to 20000.
Fig. 5. a) – Ray initialization by using an artificial triangulated spherical surface near the body; b) - ray focusing for homogeneous model (all tissues are assigned water properties); c) – ray focusing for a heterogeneous model.

Ray Propagation

Every ray carries information about acoustic amplitude decay due to

i. Transmission through boundaries;

ii. Nonlinear effects while propagating in every medium;

iii. Viscous effects in every medium;

Within every homogeneous tissue, the ray passes through without changing direction. There, the ray is subject to non-linear and viscous effects as described below. At every tissue boundary, the ray undergoes transmission and changes its direction. Reflected rays have not been considered. Fig. 5b is a zoomed in version of Fig. 5a for the homogeneous model with the caustic at the focal point. Fig. 5c is a zoomed in version of Fig. 5a for a heterogeneous model with tissue properties from Table 1.

Nonlinear Effects

Consider a ray entering a homogeneous tissue with parameters \( \rho_0, c_0 \). The corresponding pulse is a triangular shock wave of length \( l \), and with initial pressure amplitude \( P_0 \). The pulse travels distance \( \Delta x \) through the tissue. Using basic nonlinear acoustic theory it can be easily shown that the final pulse amplitude \( P_1 \) after traveling distance \( \Delta x \) is given by

Consider a ray \textit{entering} a homogeneous tissue with parameters \( \rho_0, c_0 \). The corresponding pulse is a triangular shock wave of length \( l \), and with the \textit{initial} pressure amplitude \( P_0 \). The pulse travels distance \( \Delta x \) through the tissue. Using basic nonlinear acoustic theory shown above [6,7], it can be easily shown that the final pulse amplitude \( P_1 \) after traveling distance \( \Delta x \) is given by
\[ P_1 = NP_0, \quad N = \frac{1}{1 + \frac{\Delta}{l}}, \quad \Delta = \frac{\Delta x P_0}{c_0 \rho_0} \left[ 1 + \frac{B}{2A} \right] \]

Eq. (4) is only valid for a strictly planar propagation of shock waves. The corresponding geometrical correction [11,12] may be taken into account when necessary. However, one needs to be careful with a correct decoupling of non-linear and a geometrical-convergence effects.

Viscous Damping

The data about the viscous loss factor in tissues is hardly available. We included a generic loss factor in the form \( \exp(-\alpha \Delta t) \), but haven’t used it because we haven’t found reliable sources for the \( \alpha \) coefficient for human tissues.

Geometrical Convergence

The acoustic intensity of a ray can, according to ray theory, be calculated using the principle that the power within a ray tube remains constant within that ray tube. Instead of modeling the ray tubes, we suggest to use a Voronoi diagram [14] to find the intensity distribution in every beam plane. Consider a cross-section of the focusing beam in Fig. 2c exactly at the focal point. The corresponding cross-section plane (the focal plane) is shown in Fig. 3. The dots denote ray crossing points. A polygon around every dot is its Voronoi polygon, which essentially defines an “area of influence” of this particular ray. All points within the polygon are closer to a particular ray crossing point than to any other. Even for very closely spaced ray crossing points, the Voronoi polygons create finite areas of influence as highlighted in Fig. 3 by a circle.
Fig. 6. Focal plane in Fig. 2c and the corresponding Voronoi diagram.

Now, consider a ray with an area $A$ on the Voronoi diagram in Fig. 3. This ray is already characterized by an amplitude decay factor, $D$, $D < 1$ found according to Eqs. (3) (boundary transmission loss) and (4) (nonlinear loss), respectively. Its initial area of influence was defined as $A_0$ in Section 5. The resulting field amplitude within area $A$, whether for pressure or velocity, is then given by the product of a factor

$$D \sqrt{\frac{A_0}{A}}$$

and the initial amplitude at the spherical cap in Fig. 2a. Certain local averaging variations of this method are possible to provide a more homogeneous focal field distribution.

Simulation results
Results given below should be considered as very preliminary. They were mostly used to test the algorithm and its different parts, and speed up the computational scripts.

Drift of a focal point in the focal plane

For the situation shown in Fig. 5a, the focal point will be shifted in the focal plane as compared to the idealized focusing in the homogeneous medium, for both kidneys. This is a purely geometrical effect weakly affected by nonlinearity. Fig. 4 shows the Voronoi diagram in the focal plane and the position of the focal point for the left kidney. The corresponding shift is about 3 mm.

Fig. 7. Focal plane in Fig. 2c, the corresponding Voronoi diagram, and the position of the geometrical focal point.
Figure 8. Human body model in a sphere with a point source for Gaussian wave simulations

This was the first model developed to simulate how waves would propagate through human tissue. There is a mesh for the kidney, the torso, and an ellipse that encloses both meshes. There is a point source off to the left of the middle of the ellipse. This point source is where the Gaussian explosion would be simulated.

Mesh Quality

The mesh had a maximum element size of 140 mm, and a minimum element size of 40 mm. The maximum element growth rate was 1.7, the curvature factor was 0.8, and the resolution of narrow regions was 0.3. Due to these restrictions, the Gaussian explosion was simulated with a
frequency of 10 kHz, and a tau of 0.1 ms. To have a mesh with a higher frequency would cause a wave that would be too small to be resolved on an individual element, so the simulation could not be trusted. The run time of the simulation was 2 minutes and 18 seconds, due to the large element size.

The issue with this model was the mesh quality. Due to the relatively large size of the elements, the wave cannot be resolved properly. This limited the range of frequencies that the model could be tested under. Since lithotripters use a frequency from 100 kHz to 1 MHz, we needed a better model.

Of the two main models that were created for this experiment, a high fat and low fat model, the peak pressure at each model did not seem to vary much.

The rays that are produced in the region of interest do not seem to focus perfectly. Instead of coming to a point, they scatter just a bit.
Figure 9. Compacted human body model to focus on kidney

This model is similar to the other ellipsoidal model, but on a much smaller scale. Since we are only concerned with a certain region of the torso for the lithotripter, we removed most of the torso and extra space from the model. The torso is now represented as a boundary that cuts through the middle of the ellipse. The kidney mesh is still represented by the kidney, and the rest of the model is water for the lithotripter.

Mesh Quality

The maximum element was 8 mm, the minimum element size was 0.08. The maximum element growth rate was 1.3. The curvature factor was 0.2, and the resolution of narrow edges was 1. It took 13 hours, 21 minutes, and 50 seconds to run.
This model was truncated to improve running time, so that solving the wave equation for parts of the body we weren’t interested in could be ignored. It also allowed for a greater mesh quality, so that the model could run at higher frequencies. The problem with running at higher frequencies, and having smaller element sizes, is that it took too long to run. 13 hours is not a reasonable running time for the experiment we are conducting.

Figure 10. 2D human body model

For this simulation, we attempted created a 2 dimensional axisymmetric model. While this wouldn’t be entirely representative of what would happen in 3 dimensions, it can be used to help us determine where to go next with research.
The model itself is simple, there is a rectangle that bounds the entire model. There is a curved arc near the top of the model that represents the parabolic reflector, that would be the points used as the point source for the Gaussian explosion. There is a narrow region near the middle of the model that is bounded by two horizontal lines, this represented the skin. The next boundary represented the fat, which would vary by moving the boundary near and far from the second fat layer represented by the outer circle. The inner circle was the kidney.

Mesh Quality

The maximum element size was 5 mm, and the minimum element size was 0.01 mm. The maximum element growth rate was 1.1, the curvature factor was 0.2, and the resolution of narrow regions was 1. Since the mesh was represented by small elements, it could be used with a higher frequency for the Gaussian pulse, 50 kHZ. Tau was registered at 20 ms. The run time for this model was 2 minutes and 29 seconds.

This next model was a narrower version of the 2nd model since we put it in 2d. This allowed us to set up a reasonable model for the test we wanted to run, and be able to finish the experiment relatively quickly. The problem with this model is that since it was in two dimensions, it could not accurately account for nonlinear acoustic properties. Since fat, water, and muscle have varying nonlinear parameters, we decided it would be best to develop a ray tracer to simulate our model.

Of the two main models that were created for this experiment, a high fat and low fat model, the peak pressure at each model did not seem to vary much.
The COMSOL software did not work for what we needed. We were able to use it effectively under 50KHz, but that just wasn’t reflective of the frequencies that are seen in lithotripters as they range from 100KHZ to 1MHz.

Of the two main models that were created for this experiment from the programming aspect, a high fat and low fat model, the peak pressure at each model did not seem to vary much. The rays that are produced in the region of interest do not seem to focus perfectly. Instead of coming to a point, they scatter just a bit.

Figure 11. Front view of computational human body model with reflector
Figure 12. Top view of computational human body model
Discussion

Since the high fat and low fat models did not vary too much in the peak pressure, it isn’t reasonable to say that the makeup of the tissue changes the focal area or the peak focal pressure. What may be more important is the distance from the wave source, the lithotripter, to the kidney.
stone. Since the nonlinear parameter, which determines how much the ray strength decays, is a linear function of distance you will have a lower proportion of peak pressure at the kidney. The results that show the opacity of the layers may show that the focal point isn’t at the intended focal point. This could be due to a number of things. In practice, maybe the lithotripter wasn’t aimed properly and caused the focal area to be off by a few millimeters. This shouldn’t be a problem for the computer simulation, however, because the focal point is set as the center of the sphere and the rays are given a direction that leads them to the focal point. This is supported when you make the acoustic properties of the various tissue water. In this case there would be no change in direction at the interface since they are all water, therefore there is no interface with varying acoustic properties on either side to cause the rays to divert from their intended focus. In the simulation where they are all water, there is perfect focusing that shows the targeting method is correct and accurate.

What may be causing the problem is the odd shape of the kidney, since it’s convex. This would affect the angle of the incident rays when they meet the interface of the fat and the kidney, which may make the resulting transmitted ray have an odd angle when it passes through. This situation could also occur at the interface of the kidney and the fat, where it has already gone through the kidney, and is leaving through the other side. If it offsets the transmitted angle, it would defocus the rays which could describe the results shown.
References


Appendix

Figure 15 a. High muscle cross section in the x-z plane at 54 mm in y

Figure 15 b. High muscle cross section in the x-z plane at 56 mm in y
Figure 15 c. High muscle cross section in the x-z plane at 58 mm in y

Figure 15 d. High muscle cross section in the x-z plane at 60 mm in y
Figure 15 e. High muscle cross section in the x-z plane at 62 mm in y

Figure 15f. High muscle cross section in the x-z plane at 64 mm in y
Figure 15g. High muscle cross section in the x-z plane at 66 mm in y

Figure 16a. High fat cross section in the x-z plane at 54 mm in y
Figure 16b. High fat cross section in the x-z plane at 56 mm in y

Figure 16c. High fat cross section in the x-z plane at 58 mm in y
Figure 16d. High fat cross section in the x-z plane at 60 mm in y

Figure 16e. High fat cross section in the x-z plane at 62 in y
Figure 16f. High fat cross section in the x-z plane at 64 mm in y

Figure 16g. High fat cross section in the x-z plane at 66 mm in y
Figure 17a. Pressure in Pascals from a point source
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Figure 27b. Comparison of pressure at reference point in different homogeneous body compositions