Building tools for image-guided adaptive radiotherapy of bladder cancer
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A voxel-based finite element model for the prediction of bladder deformation

Xiangfei Chai, Marcel van Herk, Maarten C. C. M. Hulshof and Arjan Bel

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Abstract

Purpose: A finite element (FE) bladder model was previously developed to predict bladder deformation caused by bladder filling change. However, two factors prevent a wide application of FE models: (1) the labor required to construct a FE model with high quality mesh and (2) long computation time needed to construct the FE model and solve the FE equations. In this work, we address these issues by constructing a low-resolution voxel-based FE bladder model directly from the binary segmentation images, and compare the accuracy and computational efficiency of the voxel-based model used to simulate bladder deformation with those of a classical FE model with a tetrahedral mesh.

Material and Methods: For ten healthy volunteers, a series of MRI scans of the pelvic region was recorded at regular intervals of 10 min over 1h. For this series of scans, the bladder volume gradually increased while rectal volume remained constant. All pelvic structures were defined from a reference image for each volunteer, including bladder wall, small bowel, prostate (male), uterus (female), rectum, pelvic bone, spine and the rest of the body. Four separate FE models were constructed from these structures: one with a tetrahedral mesh (used in previous study), one with a uniform hexahedral mesh, one with a non-uniform hexahedral mesh and one with a low-resolution non-uniform hexahedral mesh. Appropriate material properties were assigned to all structures and uniform pressure was applied to the inner bladder wall to simulate bladder deformation from urine inflow. Performance of the hexahedral meshes was evaluated against the performance of the standard tetrahedral mesh by comparing the accuracy of bladder shape prediction and computational efficiency.

Results: FE model with a hexahedral mesh can be quickly and automatically constructed. No substantial differences were observed between the simulation results of the tetrahedral mesh and hexahedral meshes (<1% difference in mean dice similarity index to manual contours and <0.02 cm difference in mean standard deviation of residual errors). The average equation solving time (without manual intervention) for the first two types of hexahedral meshes increased to 2.3 h and 2.6 h compared to the 1.1 h needed for the tetrahedral mesh, however, the low-resolution non-uniform hexahedral mesh dramatically decreased the equation solving time to 3 min without reducing accuracy.

Conclusion: Voxel-based mesh generation allows fast, automatic, and robust creation of finite element bladder models directly from binary segmentation images without user intervention. Even the low-resolution voxel-based hexahedral mesh yields comparable accuracy in bladder shape prediction and more than 20 times faster in computational speed compared to the tetrahedral mesh. This approach makes it more feasible and accessible to apply FE method to model bladder deformation in adaptive radiotherapy.
4.1 Introduction

The finite element (FE) method is a powerful computational tool for modeling soft tissue deformation. It has been widely used to simulate the mechanical deformation of soft tissues in medical surgery [84]. For radiotherapy applications, FE modeling has yielded positive results in modeling different organs such as the prostate [88;94;104], lung [105-107], bladder [108;109], liver [86] and breast [99]. FE methods produce physically plausible biomechanical tissue deformations and serve as a gold-standard in validation of other deformable registration algorithms [110;111].

In bladder cancer radiotherapy, the target and the organs at risk (e.g. rectum, small bowel, and prostate) are highly deformable structures. For instance, the uncertainty in the location of bladder tumors can be as much as 3 cm due to changes in bladder volume and adjacent organs [15;36]. Besides the day-to-day bladder deformation, significant intra-fraction bladder wall movement may occur during radiotherapy delivery [38]. So the inter- and intra-fractional anatomy variations limit the accuracy of radiation treatment and restrict the potential of dose escalation for bladder cancer [18]. In our previous paper [108], we studied bladder motion by constructing a FE biomechanical bladder model to simulate the deformation of pelvic organs caused by bladder filling change. This bladder model shows promise for the prediction of bladder deformation using just one image data set and bladder volume changes as input. It can accurately predict short-term bladder deformation and can potentially be used to improve image-guided radiotherapy for bladder cancer patients by making multiple plans from simulated artificial CT images with different bladder volumes.

However, the current FE approach used to model patient-specific bladder deformation is a very time-consuming process and requires manual labor. This process consists of two steps. The first step is model construction, in which a triangular surface mesh is extracted from a 3D binary segmentation image of each organ, exported to a mesh generator in a commercial software package Abaqus where subtraction Boolean operations are performed between overlapped surface meshes to create the new surface mesh of each organ, material properties are assigned to each organ, the new surface mesh of each organ is discretized to a solid tetrahedral volumetric mesh and finally the load and boundary conditions are applied to the surface boundary. This model construction process is done using the graphic user interface of Abaqus. Sometimes, when there are very short lines in the newly generated surface meshes, Abaqus has trouble automatically generating high quality tetrahedral meshes with reasonable resolution. In that case, manual intervention is required. The second step is solving partial differential equations, which takes another one hour for a model with 200,000 nodes and 1,200,000 tetrahedral elements. Manual user interventions and long computation time are the major roadblocks to a wide application of FE models. Reducing the labor required to generate a FE model and the time used to solve the FE model will make this method more attractive and practical for both research and clinical applications.

In theory, hexahedral meshes generally yield a little higher accuracy than tetrahedral meshes [112]. However, this only happens when both tetrahedral and hexahedral meshes are generated from the same surface meshes. In current practice, most commercial software packages are unable to automatically generate hexahedral meshes from complex geometries.
In medical imaging application, the voxel hexahedron meshing method first proposed by Keyak et al. [113] is an alternative way to generate hexahedral meshes from segmented medical image sets. By utilizing the fact that segmentation images are naturally composed of solid discretized voxels, the voxel hexahedron meshing algorithm operates by generating a regular hexahedral element from the voxel of interest in the segmentation images instead of from the surface contours. Generating volume meshes directly from the image data can eliminate the extra step of generating a 3D surface and then discretizing the solids formed by those contours. This meshing method is robust and simple and allows automatic construction of hexahedral meshes from medical images.

However, there is a trade off because the regular hexahedral elements lead to jagged or ‘digitized’ mesh surfaces. The errors introduced by this effect are particularly significant when modeling objects with relatively thin cross-sections [114]. In addition, the number of nodes generated by the voxel hexahedral meshing method can be significantly higher than that of other meshing methods. This increased node count means that the equation solving stage of the model will be computationally more expensive.

In this study, we will evaluate the extra simulation errors introduced by the jagged cubic edges and reduce the number of hexahedral elements using a uniform downsampling and multiresolution approach. The basic idea of the multiresolution mesh approach is to use a finer mesh at regions close to the load for greater accuracy and to use a coarser level of detail in other regions for computational efficiency. In the deformation simulation of the pelvic region, stress is caused by urine inflow, so the regions near the bladder were discretized with a higher resolution than the other regions.

The purpose of this paper is to develop a method of constructing voxel-based FE bladder models with a hexahedral mesh, and compare its accuracy and computational efficiency to those of a classical FE model with a tetrahedral mesh in simulating bladder deformation.

### 4.2 Materials and methods

#### 4.2.1 Data acquisition

Ten healthy volunteers (five males and five females) were investigated in this study. The median age of the volunteers was 43 years (range: 26-63). The data set in this work is the same as was used in previous papers [40;108]. Volunteers were instructed to empty their bladder and drink 300 ml of water 15 min prior to acquisition. The volunteers were positioned supine without immobilization and a pelvic phased array coil was applied. For each volunteer, a series of MRI scans of the pelvic region was recorded at regular intervals of 10 min over a period of one hour. Imaging was done on a 1.5 T system (Somatom: Siemens Medical Systems, Erlangen, Germany). The MRI parameters were: T2-weighted sequence; repetition time: 8.1 ms; echo time: 4.0 ms; reconstruction in-plane matrix: 256*256 pixels; isotropic in-plane resolution: 1.4*1.4 mm²; and section thickness: 1.4 mm.
4.2.2 Finite element model construction

Generally, construction of FE bladder model consists of two steps. First, the problem domain (human pelvic region excluding inner part of bladder) is decomposed into a large number of elements, which are connected via nodes located on their boundaries. Then the pressure load on inner bladder wall and boundary conditions are defined.

In each series of MR images, the image with a bladder volume closest to 250 cm$^3$ was selected as reference, as this was the typical bladder volume in planning images for “full bladder treatments”. For each reference image, all pelvic structures, i.e., outer bladder wall, small bowel, prostate, rectum, pelvic bone, spine and body contours were manually delineated by a single observer. Since manual contouring of the inner bladder wall was difficult on the T2-weighted MR images, the inner bladder contour was created by uniformly shrinking the outer bladder wall by 3 mm. For the hexahedral mesh models, because of the resolution limitation, the inner bladder contour was created by shrinking the outer bladder wall by 5 mm. All contours on the reference image were used to build the FE bladder model. For the other images, only the outer bladder wall was delineated for the purpose of model verification.

Four FE models with different mesh types were constructed. The first was the tetrahedral mesh adopted from our previous paper [108]. The contours on the reference image were converted to binary segmentation images and then converted into a triangular surface mesh. To maintain a reasonable volume mesh size, the mesh was smoothed and the number of triangulations describing the surface was reduced. After that, the decimated surface mesh was smoothed again. The smoothed triangular surface meshes representing the segmented organs were imported into a commercial FE method software package (Abaqus version 6.9, Dassault Systemes, Simulia BV). Then, Boolean operations between the overlapped surface meshes were done to create a hollow bladder wall and new surface meshes of surrounding pelvic organs. The inner part of the bladder becomes the boundary surface in the FE model. Due to the delineation inaccuracy, inevitably there are some overlaps between the contours of adjacent organs, like bladder with small bowel and bladder with uterus or prostate. We defined a priority score for the pelvic organs from high to low (6 to 1): bladder, uterus (or prostate), rectum, small bowel, bone and body. Boolean operations were performed based on this prioritization such that the overlapping regions between adjacent organs are always merged into the highest priority scoring organ, subsequently the surface meshes of the organs with lower priority score were modified. Organ specific material properties were then assigned to each organ. A linear elastic material model was chosen with the parameters for each organ shown in table 4.1, which was taken from literature [94;95;108]. The linear material properties were characterized by two parameters: Young’s modulus $E$ and Poisson’s ratio $\nu$ that represent stiffness and compressibility, respectively. Next, a 3D four-node tetrahedral volume mesh was created from all surface meshes using the free meshing method in the Abaqus package. The generated tetrahedral mesh filled the entire pelvic region except the inner part of the bladder. However, this mesh generator in Abaqus did not always generate high quality 3D meshes with reasonable resolution since Boolean operations between overlapped surface meshes could create new vertices that were extremely close to existing vertices. This might cause a failure of volume mesh generation or a failure of simulation due to badly shaped elements (elongated or skinny elements). When mesh generation or simulation failed, the edge of the...
segmentation image where badly shaped elements were located would be manually adjusted. Then, the surface mesh generation, Boolean operations, and volume mesh generation would be repeated on the adjusted segmentation images. During the MRI scan session, the rectal filling did not change significantly and the bladder filling was considered a stable kinematic process, so an arbitrary internal pressure of 1 kPa was uniformly applied on the inner bladder wall to simulate an arbitrary bladder filling increase. The exact bladder pressure corresponding to the known volume was unknown. However, since the model is a linear system, the deformation caused by a different pressure could be calculated by scaling results until the desired volume was reached. The initial pressure would be scaled such that the volume of the modeled bladder corresponds to the actual bladder volume. During the scans, volunteers laid on the table in supine position with MR coil attached to the belly. We therefore assume the back of the volunteer was fully attached to the table and the belly was not moveable along the anterior-posterior direction. Hence, the kinematic boundary conditions were such that the nodes on the back position were fixed in all directions and the nodes on the abdomen were restrained from displacement along the anterior-posterior direction. Sliding was not applied to the interface between different organs due to the unknown friction coefficients between pelvic organs. The whole procedure for the construction of FE model with tetrahedral mesh was done using the graphic user interface of Abaqus. More details on the development of the tetrahedral FE model have been described in an earlier paper [108].

<table>
<thead>
<tr>
<th>Organs</th>
<th>Poisson ratio (ν)</th>
<th>Young’s modulus (E)(kPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bladder wall</td>
<td>0.49</td>
<td>10</td>
</tr>
<tr>
<td>Pelvic tissue (“body”)</td>
<td>0.40</td>
<td>15</td>
</tr>
<tr>
<td>Bone (spine and pelvis)</td>
<td>0.49</td>
<td>1,000</td>
</tr>
<tr>
<td>Prostate (male)</td>
<td>0.40</td>
<td>21</td>
</tr>
<tr>
<td>Uterus (female)</td>
<td>0.49</td>
<td>20</td>
</tr>
<tr>
<td>Rectum</td>
<td>0.45</td>
<td>10</td>
</tr>
<tr>
<td>Small bowel</td>
<td>0.35</td>
<td>3</td>
</tr>
</tbody>
</table>

The second FE model with uniform hexahedral mesh, the third one with non-uniform hexahedral mesh and the forth one with low-resolution non-uniform hexahedral mesh were constructed in a similar manner. In contrast to the first model, the hexahedral meshes were directly generated from the segmentation images of pelvic structures and skipped the process of surface mesh extraction (figure 4.1). All contours on the reference image were converted into binary segmentation images with different intensities. The intensities are the same as the priority scores defined for each organ. The segmentation images of bladder, uterus (prostate), rectum, small bowel, bone and body were labeled with intensities 6, 5, 4, 3, 2, and 1, respectively. Boolean operations were performed between the binary images of different organs to label the overlap region with the same intensity as the organ with highest priority. An extra minus Boolean operation between inner and outer bladders was done to create the bladder wall. In the final segmentation image (figure 4.2), the voxels outside the body and inside the inner bladder were labeled as zero. The basic idea of the voxel meshing method is to translate each non-zero voxel in the segmentation image into a hexahedral element in the FE model and then create 8 nodes on the corners of each element. Figure 4.3 gives a 2D illustration of the translation from segmentation image to elements and nodes.
Figure 4.1: FEM construction with four different mesh types. Four models with tetrahedral, uniform hexahedral, non-uniform hexahedral and low-resolution non-uniform hexahedral meshes were created from the binary images of all pelvic structures: bony structures, small bowel, rectum, bladder wall and prostate. The binary images were first converted to triangular surface mesh. A 3D four-node tetrahedral volume mesh was then created from all surface meshes. The eight-node hexahedral meshes with uniform element size, eight-node hexahedral meshes with non-uniform element size and eight-node hexahedral meshes with low-resolution non-uniform element size were directly generated from the binary images of pelvic structures. The volume mesh filled the entire pelvic region except for the inner part of the bladder. The inner bladder and surface of body were both the boundary surface of the FE models.
Figure 4.2: The assembled segmentation image after Boolean operation between seven binary segmentation images: inner bladder, outer bladder, uterus, small bowel, rectum, pelvic bone, spine and body. Different grey values label each organ. The region outside body and inside the bladder (black color) was not translated into hexahedral elements in the FE model.

Figure 4.3: A 2D illustration of the conversion from the segmentation image to a hexahedral mesh. (a): The dark grey (green in colorful version) and light grey label the pixels of two types of tissues in the segmentation image. The pixels with black color are labeled as “zero”. The non-zero pixels in (a) are translated to elements in (b) and nodes are located on the corners of each element. The black areas in (b) are out of the boundary surfaces of FE model.

The voxel size of the original segmentation image was 1.4*1.4*1.4 mm³. To maintain a reasonable number of elements, the segmentation image was uniformly or non-uniformly down-sampled to different resolutions. In the uniform and non-uniform down-sampling process, some neighboring voxels in the segmentation image were merged into a larger voxel with unique intensity in the new image. The intensity of the new voxel was determined by the majority intensity of original voxels. When the new voxel consists of equal amount of original voxels with two different intensities, the higher intensity (higher priority score) was assigned to the new voxel. The segmentation image was first uniformly down-sampled by a factor of two. The uniform mesh was then created from the uniformly down-sampled image and it consisted of eight-node regular hexahedral elements with a uniform element size of 2.8*2.8*2.8 mm³. In the FE model, the pressure on the inner bladder wall is the only load and we are only interested in the deformation of the outer bladder wall. We therefore considered the region around the bladder wall as the region of dominance for FE simulation. Since a finer mesh in the region of dominance has a larger effect on accuracy than that in other regions, we adopted a multi-resolution strategy to create a non-uniform mesh with finer mesh in the region of dominance and coarser mesh in
all other regions. We defined the region of dominance by extending the cubic region around the bladder border by four voxel sizes in six orthogonal directions of the original segmentation image. The original segmentation image was down-sampled to two multi-resolution images. The non-uniform and low-resolution non-uniform meshes were created from these two multi-resolution segmentation images. The non-uniform mesh was an eight-node hexahedral element with eight different element sizes with a combination of 1.4 and 5.6 mm sides. The low-resolution non-uniform mesh was an eight-node hexahedral element with eight different element sizes with a combination of 2.8 and 11.2 mm sides. The element size of the two types of hexahedral mesh in the finer mesh regions around the bladder was 1.4*1.4*1.4 mm$^3$ and 2.8*2.8*2.8 mm$^3$, respectively.

![Figure 4.4](image)

Figure 4.4: For the tetrahedral mesh (a), the pressure load is applied on the smooth inner bladder surface. For the hexahedral mesh (b,c), the load is the forces applied on the nodes in jagged inner bladder wall surface such that has equivalent effect as pressure load. The yellow arrows in (b) and (c) indicate the forces in three orthogonal directions.

The generated hexahedral elements have a one-to-one correspondence to the voxels of down-sampled segmentation image. With such correspondence, the organ specific material properties were then assigned to each element. The material properties chosen for the three hexahedral mesh models were exactly the same as those for the first tetrahedral mesh model. For the hexahedral mesh, the same pressure was uniformly applied on the jagged inner bladder wall instead of on the smoothed surface for the model as was the case for tetrahedral elements (figure 4.4). The pressure distributed normal to the hexahedral element surface was converted to the force on the four nodes located on the corners of the acted upon element surface by multiplying the pressure with the element surface area. The element surfaces of inner bladder wall with pressure load were all in the finer mesh regions, so the element surface areas with load are 2.8*2.8 mm$^2$ for uniform and low resolution non-uniform hexahedral mesh and 1.4*1.4 mm$^2$ for non-uniform hexahedral mesh. Nodes were shared by adjacent element surfaces, hence the global force on each node was the summation of the force calculated from all adjacent element surfaces. For the three models with hexahedral meshes, the FE model was created by an in-house developed software and written into a text file in the format defined by Abaqus.

**4.2.3 FE simulation and volume fitting**

Once the model construction was complete, finite element analysis (FEA) was performed to determine the displacement field in the response of the pressure load by solving partial differential equations. A static direct single step was chosen for the
FEA. The analysis was carried out on a Dell with Intel Core 2 Duo CPU of 3.0 GHz, 8 GB of RAM, and 64-bit Windows XP. Single processor and 90% memory were set in Abaqus software for FEA. The model construction time and simulation time for all four FE models were recorded.

The result of FEA was a displacement vector of each volume node. This deformation vector field was transformed into the vertices of the smoothed bladder contour of the reference image, which is the same as the outer bladder surface mesh used to build the tetrahedral mesh. Each vertex of the smoothed bladder contour has a corresponding node with the same coordinates in the tetrahedral mesh, hence the deformation vector on these nodes in the model with tetrahedral mesh can be directly transformed back to their corresponding vertices of the smoothed bladder contour. For the three types of hexahedral meshes, the deformation vector map was transformed to the vertices of the smoothed bladder contour using tri-linear interpolation. Applying positive or negative scaling to these vectors allowed the construction of expanded or contracted bladders from the reference image. The scale factors were chosen such that the volume of the reconstructed bladder (enclosed by deformed outer bladder contour) was the same as the volume of the bladder in the test image. All images, including the reference image, of each MRI series were used as test images.

### 4.2.4 Accuracy of FE models

To evaluate the performance of the hexahedral meshes, we compared the accuracy of bladder shape prediction by calculating the volume and distance based errors of the three types of hexahedral meshes with those of the tetrahedral mesh.

We used the dice similarity coefficient (DSC) to measure the volume overlap between the manually delineated and predicted bladders. For two regions: manually delineated bladder $R_1$ and predicted bladder $R_2$, the DSC was defined as the ratio of the volume of intersection to the average volume

$$DSC(R_1, R_2) = \frac{\text{Volume}(R_1 \cap R_2)}{\frac{1}{2}(\text{Volume}(R_1) + \text{Volume}(R_2))} \tag{4.1}$$

Note that the volumes of $R_1$ and $R_2$ were identical because of volume fitting. The DSC had a value of 1 for perfect agreement and 0 when there was no overlap.

The local residual error was defined by computing the distance from the points in manual delineated bladder surface to the triangles in predicted bladder surface. The residual error was positive when the point from the delineated bladder surface was located outside the predicted bladder and negative when the point was located inside the predicted bladder (figure 4.5). The standard deviation (SD) of the residual errors of all nodes was calculated for bladder shape prediction of all four models.

For comparison purposes, the overlap and difference of displacement on surface nodes between the predicted bladder shapes computed using tetrahedral mesh and hexahedral meshes were also calculated.
Figure 4.5: The solid and dashed lines represent the delineated and predicted bladder surface, respectively. The points of delineated bladder located outside the predicted bladder surface are marked as circle, while the points of delineated bladder located inside the predicted bladder surface are marked as cross mark. The residual error is positive on points with a circle and negative on the points with a cross mark.

4.3 Results

The four FE models generated on average 250,000 nodes and 1,400,000 elements for tetrahedral mesh, 610,000 nodes and 590,000 elements for uniform hexahedral mesh, 730,000 nodes and 700,000 elements for non-uniform hexahedral mesh, and 100,000 nodes and 93,000 elements for low-resolution non-uniform hexahedral mesh (table 4.2). In the region of hollow bladder wall, there are on average 5,000 nodes for tetrahedral mesh, 33,000 nodes for non-uniform hexahedral mesh and 4,000 nodes for uniform and low-resolution non-uniform hexahedral mesh.

<table>
<thead>
<tr>
<th>Mesh type</th>
<th>Mean number of nodes</th>
<th>Mean number of elements</th>
<th>Mean simulation time</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tetrahedron</td>
<td>200,000</td>
<td>1,150,000</td>
<td>1.1 h</td>
</tr>
<tr>
<td>Uniform hexahedron</td>
<td>610,000</td>
<td>590,000</td>
<td>2.3 h</td>
</tr>
<tr>
<td>Non-uniform hexahedron</td>
<td>730,000</td>
<td>700,000</td>
<td>2.6 h</td>
</tr>
<tr>
<td>Low-resolution non-uniform</td>
<td>100,000</td>
<td>93,000</td>
<td>3 min</td>
</tr>
</tbody>
</table>

Constructing each FE model with tetrahedral mesh using the graphic user interface in Abaqus took on average 40 minutes for seven volunteers. This process included importing surface meshes of eight pelvic structures, doing Boolean operations between overlapped structures, assigning organ specific material properties, automatically generating tetrahedral meshes and applying pressure and boundary conditions. However, three out of ten models with tetrahedral mesh required user intervention to eliminate badly shaped elements. In these three models, part of these procedures would have to be repeated after manual adjustment of the edge of the segmentation image where badly shaped elements were located. Constructing the FE model with hexahedral mesh required less than 1 min.

The mean simulation time of all four FE models with tetrahedral, uniform hexahedral, non-uniform hexahedral and low-resolution non-uniform hexahedral meshes was 1.1 h, 2.3 h, 2.6 h and 3 min, respectively.
The numerical simulation results from the four types of meshes showed highly similar bladder deformation patterns (figure 4.6) and produced similar predicted bladder shape (figure 4.7). The deformed bladder contours in figure 4.6 computed using the tetrahedral meshes were smoother than those computed using the hexahedral meshes.

![Simulation results of the deformed bladder wall of volunteer 2](image)

**Figure 4.6:** Simulation results of the deformed bladder wall of volunteer 2, with tetrahedral (a), uniform hexahedral, non-uniform hexahedral (c) and low-resolution non-uniform hexahedral (d) mesh. The inner bladder wall is the boundary of the FE model.

![MR scans of volunteer 4](image)

**Figure 4.7:** (a) is the 3rd MR scan (reference) of the volunteer 4 in sagittal direction overlaid with the bladder contour (thin solid line). The bladder contour in (a) is used to build the FE model. (b) (c) (d) and (e) are the 7th MR scans of volunteer 4 in sagittal direction overlaid with contours of actual bladder and predicted bladder. The thin solid contour around the black regions is bladder delineation of 7th MR scan. The thick solid contours are the predicted bladder shapes computed by tetrahedral mesh (b) uniform hexahedral mesh (c) non-uniform hexahedral mesh (d) and low-resolution non-uniform hexahedral mesh (e), respectively.

The overlap index DSC and the SD of residual error between the predicted and actual bladder shape as a function of bladder volume change for a sampled volunteer 4 was plotted in figure 4.8. The DSC decreased and SD of residual error increased with increasing bladder volume change. For every MR scan, the predicted bladder shape computed using the tetrahedral mesh always had slightly higher volume overlap and lower residual error with the real bladder shape than those computed using the three types of hexahedral meshes (figure 4.8). In figure 4.8, the overlap index was not one and SD of residual error was not zero at zero volume difference. This was because of the smoothing effect and volume fitting error.
Figure 4.8: The DSC overlap index (a) and the SD of residual error (b) between predicted and actual bladder contours versus bladder volume change for volunteer 4. The triangular, circle, diamond and square icons indicate the results computed by tetrahedral mesh, uniform hexahedral mesh, non-uniform hexahedral mesh and low-resolution hexahedral mesh, respectively.

Figure 4.9 and figure 4.10 summarize the mean DSC and mean SD of residual errors for each volunteer derived by the four models with different mesh types. Since the residual errors are a signed distance and the predicted bladder has the same volume as the real bladder, the mean residual errors for the four models were all almost zero (<0.01 cm). For all volunteers, the mean DSC and mean SD of residual error (table 4.3) both showed that the predicted bladder shape computed by the three types of hexahedral mesh had just slightly lower conformity with the real bladder than those computed by the tetrahedral mesh (less than 1% reduction of mean DSC and less than 0.02 cm increase of mean SD of residual error). There was no significant difference in the DSC and residual error between the predicted bladder shape computed using the three types of hexahedral meshes. The overlap between the bladder volumes reconstructed by the tetrahedral mesh and the three types of hexahedral meshes were 0.961 (uniform hexahedral mesh), 0.959 (non-uniform hexahedral mesh) and 0.951 (low-resolution non-uniform hexahedral mesh), respectively. The SD of displacement difference between the reconstructed bladder shape computed using the tetrahedral
mesh and the three types of hexahedral meshes were 0.154 cm (uniform hexahedral mesh) and 0.161 cm (non-uniform hexahedral mesh) and 0.212 cm (low-resolution non-uniform hexahedral mesh).

![Figure 4.9: The mean DSC volume overlap index between predicted and actual bladder contours for ten volunteers.](image)

![Figure 4.10: The mean standard deviation of local residual error between predicted and actual bladder contours for ten volunteers.](image)

**Table 4.3: Mean DSC and mean SD of local residual error over all volunteers by four bladder prediction models.**

<table>
<thead>
<tr>
<th>Mesh type</th>
<th>Mean DSC</th>
<th>SD residual error (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tetrahedron</td>
<td>0.872</td>
<td>0.366</td>
</tr>
<tr>
<td>Uniform hexahedron</td>
<td>0.865</td>
<td>0.380</td>
</tr>
<tr>
<td>Non-uniform hexahedron</td>
<td>0.866</td>
<td>0.377</td>
</tr>
<tr>
<td>Low-resolution non-uniform hexahedron</td>
<td>0.866</td>
<td>0.378</td>
</tr>
</tbody>
</table>
4.4 Discussion

We have presented a voxel-based finite element bladder model, using all-hexahedral meshes that can be constructed directly from segmented medical images. The models that use a hexahedral mesh have comparable accuracy in bladder shape prediction as the standard model with tetrahedral mesh (overlap decreased <1% and SD of residual error increased < 0.02 cm). Using the low-resolution hexahedral mesh, the simulation time can be dramatically reduced without losing prediction accuracy.

We observe that the deformed bladder wall computed using the tetrahedral meshes is slightly smoother (figure 4.6) and slightly closer to the real bladder shape (figure 4.7) than those computed using the hexahedral meshes. This is primarily due to the extra simulation error and numerical instability introduced by the jagged cubic edges between the different structures. The results computed using the three types of hexahedral meshes are comparable. The larger ‘step’ edge effect in the low-resolution non-uniform hexahedral mesh does not yield a significant decrease in prediction accuracy. However, since the surfaces of human organs are supposed to be smooth, these edge effects can possibly be reduced by applying certain smoothing filters on the mesh surfaces [115]. Another reason for the lower accuracy in hexahedral meshes is that the thicker bladder wall and different organ shapes defined in the models with hexahedral meshes can affect the bladder deformation pattern and as a result also affect the prediction accuracy. To study the effect of thicker bladder wall, we did a limited analysis to test how the variation of bladder wall thickness affects the accuracy of the bladder model. For volunteer 4, besides the tetrahedral model with 3 mm bladder wall, we built another two tetrahedral models with bladder wall thickness 5 mm and 10 mm (figure 4.11 (a), (b) and (c)). We used three models with different bladder wall thickness to predict the bladder shape of the 7th MR scan of volunteer 4 with about 250 ml bladder volume increase. The two models with thicker bladder wall in figure 4.11 (e) and (f) show very similar deformation patterns as the one with 3 mm bladder wall in figure 4.11 (d). The overlap between the predicted bladders reconstructed by the models with 3 mm bladder wall and other two models with thicker bladder wall is 0.989 (5 mm bladder wall) and 0.978 (10 mm bladder wall), respectively. In this particular case, the two models with thicker bladder wall even produced a little higher prediction accuracy than the model with 3 mm bladder wall (0.002 and 0.005 DSC increment for the models with 5 mm and 10 mm bladder wall). This suggests that the variation of bladder wall thickness has only little influence on the accuracy of bladder model, i.e. the shape of the deformed bladder is greatly determined by the surrounding organs. In addition, the material properties of some organs were previously obtained by optimizing on a subset of volunteers based on the tetrahedral mesh [108]. Therefore, these material properties are not the optimal values for hexahedral meshes. But the extra errors introduced by all of these effects are very limited (SD of residual error increment <0.02 cm). This extra residual error increment is negligible, compared with the mean SD of residual simulation error (0.37 cm).
Figure 4.11: Comparison of bladder deformation simulation with three different bladder wall thickness. Tetrahedral FE models of volunteer 4 with three different bladder wall thickness 3 mm (a), 5 mm (b) and 10 mm (c) are separately constructed. (d), (e) and (f) are the simulation results with same bladder volume from model (a), (b) and (c), respectively.

Compared to other volume mesh generation approaches, the voxel-based mesh is more robust for complicated structures. In most literatures about FE model for radiotherapy application [86;106-108], the contour delineation of each structure must first be converted into a binary segmentation image, then a triangular surface mesh, and eventually a tetrahedral volume mesh. In practice, decimation and smoothing of the surface mesh is applied to keep a reasonable number of volume mesh and to grossly simplify the model geometry. The surface smoothing usually can only preserve the shape of some structures with simple topology, e.g. prostate, bladder, lung, or breast. However, for more complex structures like the lung bronchial tree and fibro glandular breast tissue, because of the disconnectedness of the structure and delineation errors, the shape of these structures cannot be described by a single closed surface mesh. In contrast, the voxel-based method can fully translate any binary segmentation to FE model without geometrical limitations. Especially with the MR elastography technique [102;103], which is able to measure the material properties in vivo, assigning the measured material properties directly to each individual element instead of to the bulk of the structure could result in a more accurate simulation without requiring segmentation.

The accuracy of the bladder model has a high correlation to the magnitude of the bladder volume as shown in figure 4.8. The maximum local residual error over all volunteers was 2.65 cm, as the bladder volume change reaches up to 500 ml. Such a large residual error is not clinically acceptable. Within 100 ml bladder volume difference, the maximum local residual error is always smaller than 1 cm. Pinkawa et al. [77] reported that the SD values of daily bladder volume are 124 ml with full bladder protocol and 56 ml with empty bladder protocol. The bladder volume increase during a 28 min treatment time is $42 \pm 47$ ml. Two possible clinical applications of FE based bladder model is to predict bladder filling during treatment and create multiple
plans with different bladder volume for an online ART with empty bladder protocol. In our study, the average bladder volume difference is 198.8 ml, which is larger than the intra- and inter-fractional bladder volume variations reported above. So, a higher accuracy could be achieved when used in clinical applications.

The image data and tetrahedral models used in this paper are the same as that in the previous study [108]. However, the average value of DSC for tetrahedral mesh is lower than that in the previous work by 0.017. This is because in the previous study a smoothed bladder contour from the test image was used to calculate DSC overlap, while in the current study, a binary segmentation image without smoothing from the test image was used to calculate the DSC.

The computation time of solving FE equations is highly dependent on the number of nodes and elements and also the size of physical memory. During simulation, Abaqus/Standard software creates two groups of temporary data. The first group of data must be stored in memory and the second group of data can be stored either in memory or on hard disk. In our study, the models with tetrahedral mesh and the first two types of hexahedral meshes all require the use of hard disks to store temporary data and also require long computation times. Using multiple CPU cores will not help speed up the computation as most of the computation time is spent on transferring data between memory and hard disk. For the model with low-resolution non-uniform mesh, the two groups of temporary data can both be stored in memory. Since accessing data in memory is much faster than accessing data on hard disk, the computation time of the model with low-resolution non-uniform mesh is dramatically reduced to 3 min. In general, using additional memory or faster hard disks (e.g. solid state drives) will often speed up the computation time more than using a faster or multiple-cores CPU.

The limitations of the voxel-based mesh are such that it can create only solid elements but not shell or beam elements and that it cannot include sliding in simulations because of the jagged interface between organs. Applying smoothing filters on the mesh surfaces between different structures [115] might allow a simulation with sliding.

The voxel-based FE model significantly simplifies the model construction process. By skipping the solid mesh generation process and using a non-uniform downsampling approach, a lot of user labor and simulation time can be saved in the FE modeling process. As the computational power of technology increases, the FE model will become more accessible in clinical applications.

For whole bladder radiotherapy, the target is highly deformable between and during treatment fractions due to bladder filling. So, only translation correction in IGRT is usually not sufficient to compensate the deformable motion. Recently, various groups have developed online ART strategies for bladder cancer [22;23;55;58]. In these treatment strategies, multiple plans with different bladder volumes are created and subsequently, the smallest plan safely covering the bladder on the CBCT is selected as the plan of the day. Online ART can improve target coverage or reduce the amount of small bowel irradiated [22;23;55;58]. The multiple plans with different bladder volumes are created from the planning CT and extra images acquired before treatment or during early treatment. Using our method, we can easily build a FE model of the
pelvic region from one planning image, quickly simulate the deformation of the pelvic organ caused by bladder filling change, create multiple artificial CT images with different bladder volumes, and make multiple plans for the artificial CT images. This mechanical deformation simulation can greatly facilitate the practical implementation of adaptive radiotherapy procedures by saving extra CT scans. The FE model can also be used to segment the bladder on images acquired during IGRT.

Manual segmentation of pelvic structures is a time-consuming process, automatic or semi-automatic segmentation is always desired to reduce the manual workload. However, in practice for bladder cancer patients, the contours of bladder, body, and some organs at risk (small bowel and rectum), already exist for treatment planning. Bony structures can be simply segmented by applying a thresholding filter to planning CT. Only a little extra workload is therefore needed to obtain the extra segmentation of pelvic structures as inputs to the FE bladder model.

4.5 Conclusion

Voxel-based mesh generation allows fast, automatic, and robust creation of finite element bladder models directly from binary segmentation images without user intervention. Even the low-resolution voxel-based hexahedral meshes yield comparable accuracy in bladder shape prediction and more than 20 times faster in computational speed compared to the tetrahedral mesh. This approach makes it more feasible and accessible to apply FE method to model bladder deformation in adaptive radiotherapy.