PREDICTION OF PROSTATE MOTION AND DEFORMATION USING FEM MODELING FOR BETTER BIOPSY ACCURACY

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Abstract. Registration between MRI and TRUS volumes is essential in prostate biopsy or brachytherapy for physicians to prepare targeting plan of tumor sites before needle insertion. However, prostate motion induced by needle puncture has been observed during the clinical procedure, which increases the needle targeting error and lowers down the targeting accuracy attained by the registration progress. In this paper a 3D pelvic model was constructed using ABAQUS based on the medical images. The 3D FE modeling establishes a predictive correspondence between an input needle puncture action and the induced prostate motions. Quantitative analysis of prostate motion and evaluation of the methods to reduce prostate motion upon needle puncture were carried out in terms of equivalent rigid body motion. It is found that the reduction of prostate motion is limited through the way of probe sheath push towards prostate for the purpose of reinforcing prostate's constraint. However, increasing needle insertion speed was demonstrated the effective way. The paper also discussed the details on how to model surrounding tissues and the cohesive interconnections between organs for better prediction of prostate motion during biopsy. The study provided insight into the key factors that determine the prostate motion and deformation. The nonlinear biomechanical FE models will be used to fusion two types of medical images and give guideline for real-time operations to further improve the tissue sampling accuracy.

1 INTRODUCTION

Prostate cancer is a major health threat to aged men in the world. Magnetic resonance (MR) scanning is a non-invasive imaging modality and has been used to produce clear sectional images of pelvic organs for identification of prostate tumors. However, the localization of tumors by MR technique is not determinate and some researches were focused
on improvement of its detection accuracy [1,2]. Therefore, after MR scanning and localizing the suspicious tumor sites in prostate gland, needle biopsy is still required in hospital for histological verification and characterization of prostate cancer. Currently, the procedure of MR-guided needle biopsy is still a technical challenging in practice [3,4], and transrectal ultrasound (TRUS) imaging is the routine clinical standard method for real time guidance to needle biopsy and prostate cancer treatment. TRUS guidance is simple in operation and also inexpensive, yet the imaging is too blurry to identify tumor sites expect for a rough prostate contour. Although elastography has been a recent research topic to improve tissue identification and characterization from TRUS imaging, its application still requires further clinical evaluation [5,6]. To date, the practical approach in hospital is using MR imaging to target the suspicious regions of prostate tumors prior to biopsy or brachytherapy procedure, and using TRUS imaging to guide the needle toward the targets during the procedure. Due to prostate motion and deformation arising from MR endorectal coil and TRUS probe interventions as well as patient’s different gestures, registration to establish the correspondence between MR and TRUS prostate volumes is an essential but complicate process for this MR-targeted TRUS-guided approach [7-12].

However, physicians had found that the actual target points after needle insertion did not match the planned locations before needle insertion. For example, prostate rotation can reach up to 13.8° in the coronal plane during prostate brachytherapy [13], and the average biopsy targeting error reaches around 5.4 mm [14]. Prostate motion and deformation caused by needle puncture force (up to 2.5 N at capsule rupture, [15]) should contribute to this targeting error. A method for needle to be tapped instead of pushing into prostate was developed; and experimental measurements showed the mean prostate motion reduced significantly from 5.6 mm (range 0.3–21.6 mm) to 0.9 mm (range 0–2 mm) as can be found from [16].

In this paper, the prostate motion and deformation upon biopsy intervention were studied based on finite element (FE) modelling. The 3D pelvic FE model was constructed completely within ABAQUS/CAE, and the different parts with analytical geometries were created for bladder, prostate and rectum from MR images. Instead of described by simple linear elastic material, the hyperelastic material properties from experimental data were derived and assigned to these three organs. The surrounding tissues which their effect on three organs motion cannot be omitted were also considered and simplified as a linear elastic body. Cohesive interaction was used to simulate the traction-separation behavior between organs. Interaction of fluid cavity was applied to the shell bladder, to simulate the internal urinal pressure of bladder when compressed. With this 3D model, the prostate motion and deformation upon biopsy intervention were predicted, and characterized in terms of equivalent rigid body motion. The simulations explored and evaluated some most possible methods to reduce prostate motion during needle insertion; these include reinforcing prostate constraint by probe sheath push, and the dynamic needle insertion procedure.

2 FINITE ELEMENT MODEL OF TRUS BIOPSY

2.1 Creation of FE parts for pelvic organs

Two-dimensional FE pelvic model was constructed [17] using the open source software to describe quantitatively the effects of organ geometries and boundary conditions on prostate motion. Three-dimensional FE pelvic model was also built up [18] by software CATIA to
study prostate motion caused by bladder and rectum filling. As organs are geometrically irregular, construction of their FE parts from medical scanning images is challenging; even with the help of some software to convert segmented images to FE meshes (e.g. the open source software: MeshLab, Visual Computing Lab). Therefore, in this paper we introduce the method to build up the 3D FE pelvic model completely through ABAQUS (version 6.12-2).

To create the FE parts for the pelvic organs, the contours of the cross-sections were first delineated from the MR images in a series of datum planes that represent the cross-section positions. Calibration of scales should be made in this stage to make dimensions of FE part coincide with MR scanning. Then the solid and shell LOFT feature (in ABAQUS/CAE part module) were applied through these closed spine wires respectively to create the 3D solid part for prostate and shell parts for bladder and rectum (Fig. 1a). By this LOFT method, all the parts were created with analytical geometry. In this study, the global coordinate of FE model follows the MR scanner, where XY plane denotes axial plane, YZ plane denotes sagittal plane and XZ plane denotes coronal plane.

As the spine wires were delineated from the outer surface of the bladder and rectum wall, the loft surfaces were defined as top surface of the shell sections. The average value of 5 mm and 3 mm shell thickness were assigned to the bladder and rectum respectively. The dimensions and locations of the FE parts retained the same coordinate values as MRI scanning, so that the instances generated in assembly were positioned automatically as their anatomic relationship, as shown in Fig. 1b. The bladder and rectum shells were meshed by the 3-node triangular general-purpose shell elements (S3), and the prostate was meshed by the 4-node linear tetrahedron elements (C3D4). Global element control size of 3 mm was applied to the three organs. The probe sheath part (shaft length 145 mm with a φ20 mm hemispherical front end) was created as an analytical rigid shell (element type ARSR) and positioned in place to insert into the rectum.

All other organs and tissues which surround the bladder, prostate and rectum (the surroundings) were also included in the 3D model, and simplified as a spherical elastic body.

Figure 1: (a) construction of the 3D FE parts from cross-section contours of pelvic organs by the LOFT method; (b) assembly of the bladder, prostate and rectum instances for TRUS probe intervention; (c) the part of surroundings created by CUT method.
The feature of “merge/cut instances” in ABAQUS/CAE assembly module was used to cut out the inner void region from the solid spherical body by the bladder/prostate/rectum assembly, as shown in Fig. 1c. The created part of surroundings part also has analytical geometry and can be re-meshed conveniently in ABAQUS. The part of surroundings was meshed with tetrahedron elements (C3D4); the global element size is 10 mm and its inner surface is assigned by 3 mm element size. To ensure the size of the surrounding body is large enough so that movements of the inside organs will generate negligible effects on its outer surface (which is fixed in simulation), sphere diameter of 200 mm was used and considered reasonable in view of the anatomic size of pelvic region.

2.2 Material properties, interaction properties and boundary constraints

Considering the organs may experience large strain upon probe intervention and push, nonlinear elastic properties are used in the current 3D model. However, mechanical tests of pelvic tissues up to large deformation are quite limited and scarce in literatures. Fig. 4 shows the nonlinear stress-strain curves from literatures which were implemented in current model. The curves of rectum and bladder were measured from uniaxial tensile tests [19,20], and were described by Ogden model of hyperelasticity [18]. The curve of prostate was the average response of uniaxial tensile testing on prostate tissue slices [21]. For the part of surroundings, linear elasticity was often adopted and to our knowledge there is no non-linear description in literatures. Therefore, the Young’s modulus of 15 kPa was taken in current model, referring to the literature [22]. The Possion’s ratios of all the pelvic tissues were also referred to this literature: a high value of 0.499 was assigned to bladder and rectum and a low value of 0.4 was assigned to prostate and surrounding bodies. Hyperelastic Marlow model was employed in current 3D model for all three organs, and was defined by the corresponding stress-strain curves.

![Figure 2: The stress-strain curves from experimental testing of tissues from rectum and bladder [19,20] and prostate [21].](image)

The current pelvic model includes four organ parts (bladder, prostate, rectum and surroundings) and a probe sheath. Interaction properties were created to describe the contact behaviors between these parts as well as the fluid action in bladder. Tangential “frictionless” and normal “hard” contact relationships were defined to simulate the compressional interactions between the four organ parts, with surface-to-surface contact formulation and penalty constraint enforcement method. The probe sheath only has interactions with the inner
surface of rectum, and was assigned the frictionless-hard compressional contact with node-to-
surface contact formulation.

Besides of the compressional interactions between pelvic organs, there are some degrees of
connection forces from the inter-organ tissues such as membranes etc. The surface-based
cohesive behavior assumes a linear elastic traction-separation law prior to damage, and its
uncoupled format was used in current model, where the nominal traction stress vector \( \mathbf{t} \) is
related to the separations vector \( \mathbf{\delta} \) by the elasticity matrix \( \mathbf{K} \) (subscript \( n \) denotes normal
direction; \( s \) and \( t \) denotes shear directions),
\[
\mathbf{t} = \begin{bmatrix} t_n \\ t_s \\ t_t \end{bmatrix} = \begin{bmatrix} K_{nn} & 0 & 0 \\ 0 & K_{ss} & 0 \\ 0 & 0 & K_{tt} \end{bmatrix} \begin{bmatrix} \delta_n \\ \delta_s \\ \delta_t \end{bmatrix} = \mathbf{K}\mathbf{\delta} \tag{1}
\]

Experimental data on the normal and shear stiffness are not available from literature for the
pelvic organs’ interactions. Therefore, approximation values (\( K_{nn} = 10K_{ss} = 10K_{tt} = 0.001 \text{MPa/mm} \)) were used to demonstrate the implementation and study the effects of
cohesive connection between the inner surface of surrounding body and the outer surfaces of
the three organs. For the contact region between bladder and prostate, it is assumed there are
joint tissue connections, so that the penalty contact enforcement method was employed in the
cohesive interaction definition to reinforce the connection in both normal and tangential
directions.

In the current 3D model, the bladder was modeled by a shell capsule full filled with urine,
instead of a solid body or a capsule assigned with constant internal pressure. This modelling
approach for bladder will improve the fidelity of mechanical response of bladder - the internal
liquid pressure will increase upon external push on bladder, which will in turn increase the
overall stiffness of the bladder. In ABAQUS, the interaction property of surface-based fluid
cavity can model the mechanical response of liquid-filled structure; it supersedes the element-
based hydrostatic fluid cavity capability in functionality without the need to define fluid or
fluid link elements. The fluid inside a cavity is modelled with its volume \( V \) as a function of
the fluid pressure \( p \), temperature \( \theta \) and mass \( m \) (\( V = V(p, \theta, m) \), ABAQUS theory manual).
This function meets the primary requirement of current 3D model. Therefore, the surface-
based fluid cavity was created and associated with a reference node defined in the bladder.
The cavity reference node can output the variation of fluid pressure and volume inside the
bladder.

The portion of urethra that connects to prostate is not included in current 3D model. To
include the effect of its tractions to prostate when prostate motion occurs, linear springs
(element type SPRING1) are assigned to the nodes on prostate left end face and connect to
ground along global \( Z \) axis to simulate the remote traction direction. An approximation is
made that 1 mm extension results in 1 N traction force on prostate end.

Finally, boundary conditions were applied to the model to represent the actual anatomic
constraints, as shown in Fig. 1. To facilitate the insertion of probe sheath into rectum, nodal
rotation of whole rectum wall and translational displacement along \( Z \) direction at rectum left
end are prohibited (namely for rectum wall \( UR1=UR2=UR3=0 \), rectum left end \( U3=0 \)). The
public bone is not included in this model, so that its constraint on bladder is represented by
fixing the movements of bladder upper portion in \( Y \) and \( Z \) directions (\( U2=U3=0 \)). The outer
surface of surrounding body is also set fixed for all the active degrees of freedom (U1=U2=U3=0), by assuming that it will have negligible effect on internal deformation.

3 SIMULATION RESULTS OF PROSTATE MOTIONS

3.1 Pelvic organs’ behaviors upon probe sheath insertion

During the TRUS biopsy or brachytherapy procedure, the probe sheath is inserted into rectum to appropriate depth beneath the prostate and turn towards it to a certain position, after that the sheath is fixed and ultrasonic scanning and needle puncture proceed. Therefore, besides of the initial step in ABAQUS, three steps are set up in the 3D model to simulate this process. As shown in Fig. 3a, the reference point of rigid probe sheath part is first assigned a displacement U3 along -Z direction. As the total length of the sheath is 145mm in this model (equal to the maximum insertion depth of the currently employed biopsy device), the reference point reaches around the anus position after this step. Then, an anticlockwise rotation is assigned to the reference point to turn the sheath front end upwards (up to 5°) to push the prostate. General static procedure of ABAQUS standard is used for these two steps to simulate the quasi-static process. In the third step, a concentrated force is applied to the prostate to simulate prostate motion upon needle insertion before capsule rupture; depending on the needle speed, general static or dynamic implicit analysis was applied respectively for this step.

Figure 3: (a) The assembly of 3D pelvic model for biopsy procedure (in the figure, the probe sheath has been inserted horizontally into the rectum and positioned at an upward rotation of 5°); (b) the corresponding stress and deformation in prostate cross-section.

From simulation results when the probe sheath is upward rotated at 5°, the von Mises stress is within the range of 10^{-4} to 10^{-5} MPa at surroundings outer surface; the contact force magnitude increases with the sheath rotation and reaches up to 3.3 N between prostate and bladder, and 14.7 N between prostate and rectum; the total force applied on the probe sheath is mainly in the Y direction and reaches up to 27 N; and the internal fluid pressure in bladder increases with the compression on bladder and reaches up to 2.2 kPa. The deformation and distributions of stress are quite not uniform within the prostate as shown in Fig. 3b.

3.2 Characterization of prostate motion upon needle puncture

It was reported [17] that the maximum prostate deformation occurs during the pre-rupture phase based on observations of needle insertion process. A displacement boundary condition
of 3.25 mm was applied to a node of prostate to study the prostate motion up to capsule rupture. The load of concentrated force is linearly ramped up with time and orientated along global Z direction to represent the long needle shaft.

To characterize the prostate motion and deformation, we used the concept of equivalent rigid body to evaluate the overall prostate behavior upon needle puncture. Both translational displacements (UC) and angular displacements (URC) of prostate centroid were output. The component variables 1, 2, and 3 refer to global directions X, Y, and Z, respectively.

Fig. 4 shows the displacement components of prostate centroid for a needle force up to 5 N. Besides, a general static procedure was adopted to simulate the case that the dynamic effect is totally ignored. The prostate rotation around X axis (URC1) is dominant when needle acts on the upper point. However, the needle puncture at side point generates not only large rotation around Y axis (URC2), but also obvious rotation around Z axis (URC3).

An examination of mesh size effect was also performed as shown in Fig. 4. Global element size of 4 mm, 3 mm or 2 mm was applied to each model for the parts of bladder, prostate, rectum and the inner surface of surrounding body. It demonstrates that all these mesh size can produce outputs of very good agreement for the selected parameters. Therefore, the model with 3 mm mesh size was chosen for all the following simulations in this study.

![Figure 4: Angular and translational displacement components of prostate centroid by needle applied force. (a) angular displacement for needle at upper point; (b) translational displacement for needle at upper point; (c) angular displacement for needle at side point; and (d) translational displacement for needle at side point.](image)

### 3.3 Effects of prostate constraint and needle speed

In order to reduce the targeting error caused by prostate motion during needle puncture, the effects of increasing prostate constraint and needle speed were explored and simulated in this
section. To facilitate evaluation, angular displacement magnitude (URCM) and translational displacement magnitude (UCM) were used instead of their three components which from the prostate equivalent rigid body.

To verify the effects of prostate constraint, simulations were conducted by applying a needle puncture load (linear ramp up to 5 N) on upper point of prostate with the probe sheath rotates at 0° (initial position) and 5°, respectively. Quasi static and dynamic implicit analysis (ABAQUS/Standard) were carried out to examine prostate motion upon different needle insertion speed. For dynamic simulation, the needle load was linearly ramped up to 5 N with different durations (100 ms, 50 ms, 25 ms, 17.5 ms and 10 ms) to simulate effect of dynamic process. Fig 5 showed that the reduction of prostate motion by increasing its constraint was quite limit; and the effect is not consistent. However, the reduction tendency is consistent with the increase of needle loading speed. The reduction of prostate motion is just effected from loading duration of 50 ms. Considering the approximate wave speed in prostate (4.5 ~ 7.7 m/s, estimated from an elastic modulus of 20 ~ 60 kPa), the 50 ms duration corresponds to around 3~5 rounds of wave propagation in a prostate size of 4 cm – the start of uniformity of dynamic stress. Therefore a shorter needle loading duration before capsule rupture is the key to utilize the dynamic inertial response of prostate in order to reduce its motion.

![Figure 5](image)

**Figure 5**: Angular and translational displacement magnitude of prostate centroid at different needle loading rates. (a) angular displacement with probe at 0°; (b) translational displacement with probe at 0°; (c) angular displacement with probe at 5°; and (d) translational displacement with probe at 5°.

It is interesting to find that discrepancy between the two cases (probe at 0° and 5°) at same loading rate decreases with the increase of loading rates. At the rate of 5 N/10ms, the angular and translational displacements of the two cases are almost same. This demonstrates that beyond a certain needle speed, the overall behavior of prostate is insensitive to the degrees of
constraint force to the prostate. For this reason, it is preferable to apply the needle insertion to a prostate which is free from probe push force; this can cause less prostate deformation before needled insertion so that the complexity of registration is relieved and its accuracy can be improved.

4. Discussion and conclusion

To improve the needle targeting accuracy in prostate biopsy or brachytherapy, registration between MRI and TRUS volumes to locate tumors on TRUS volume is essential for physicians to plan the targeting sites before needle insertion. However, prostate motion induced by needle puncture force may greatly lower down the possible high targeting accuracy established by the complex registration progress. Therefore, it is crucial to limit prostate motion which has been observed by clinicians. This paper established a 3D FE model to simulate the pelvic organs’ motion during the biopsy intervention procedure. It introduced the approach in ABAQUS to create geometrical irregular parts for the pelvic organs, and the interaction and contact properties in assembly to attain anatomic fidelity.

The 3D model comprises the three major organs (bladder, prostate and rectum) and its surrounding body. Though the modulus of surrounding tissue reported can varied between 3.25~30 kPa, we want to point out that the inclusion of the surrounding body in FE model is crucial for the correct output of prostate motion. As shown in Fig. 6, the prostate motion predicted are very close for 15 kPa to 30 kPa modulus; though there is somewhat deviation if the modulus is 3.25 kPa which is the lowest value found in literature. However, if the surrounding body is not included (as the assembly shown in Fig.2), the prediction of prostate motion upon needle insertion is basically unbelievable.

Cohesive interaction describes the traction separation behavior between parts. Comparisons were made between cohesive connection and tie connection as shown in Fig. 7, It shows that the tie constraint produced a little stiffer interconnections (hence less prostate motion) when compared to the cohesive results with \( K_{\text{in}} = 1.0 \) MPa/m. The figure also illustrates the results with the cohesive stiffness ten times less or larger than the above value; the corresponding prostate motion also increases or decreases accordingly. This demonstrates
that the cohesive interconnection forces between pelvic organs are essentially important for accurate prediction of prostate motion.

Using the constructed model, investigations were carried out to evaluate some possible clinical approaches to reduce target errors upon needle biopsy. Simulation results revealed that pushing of prostate by probe could not reduce prostate motion significantly though it could increase the constraint forces to prostate. On the other hand, increasing biopsy needle speed could be an effective way to for this purpose. It was shown that beyond a certain needle speed, the effect of reinforced constraint by pushing prostate is negligible. The finding is instructive because no probe pushing is needed if high speed needle insertion employed, which also helps simplify registration process.

Finally we want to point out that in the dynamic implicit modelling for high speed needle puncture, the material properties for organs are not modeled as rate sensitivity. The post-rupture process after needle penetration is not considered, by assuming that prostate post-rupture motion is smaller comparing to the magnitudes of pre-rupture motion. Future work may employ rate sensitive material properties in dynamic modeling, and examine the prostate post-rupture motion through much more complex and expensive simulations of needle penetration process.

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