Rapport de stage d’option scientifique

A Mechanical Model For the Opening of the Female Urethra

Non confidentiel
Abstract

Urinary incontinence is a common problem affecting millions of women worldwide. The successful treatment of these patients will depend on accurate assessment and diagnosis. Reliable diagnosis of urethral dysfunction requires a detailed understanding of the interactions between all the factors that control and affect the behaviour of the urethral tube. In this report, a three dimensional finite element model of the female urethral tube is presented. Its goal is to validate or rebut a new theory by a surgeon, Professor Petros, stating that muscle forces acting on the vagina contribute substantially to the process of micturition.

After briefly introducing the recent theory on the opening of the female urethral tube and the anatomy and properties of the urethral tube, we present our three dimensional model. The model takes into account the detailed composite structure of the tube, the internal fluid pressure, external constraints and external active opening forces. The mechanism of opening of the female urethra during micturition appears to be a complex combination of internal pressure generated by the contraction of the bladder and the action of external forces. We focus on the three dimensional geometry of the open tube resulting from the interactions between internal pressure, tube structure, material properties and external constraints and forces.

It is shown that accurate simulation of the dimensions and shape of the open tube is only possible when all of the above factors are taken into account. We show finally that the model is capable of simulating the complex geometry of the urethral tube observed in X-Ray photographs.

The model can be used in the near future to study the effects of tissue properties and various forms of dysfunctions on the micturition process.

Résumé

L'incontinence urinaire est un souci commun à des millions de femmes a travers le monde. Le traitement efficace de ces patientes dépend de la qualité des connaissances anatomiques et des diagnostiques. Un diagnostique fiable d’un dysfonctionnement au niveau de l’urètre ne peut exister que par une compréhension détaillée des interactions entre tous les facteurs qui contrôlent et affectent le comportement de l’urètre. Dans ce rapport, nous présentons un modèle en trois dimensions de l’urètre féminin. L’objectif est de valider ou de réfuter la théorie d’un chirurgien, Professeur Petros, selon laquelle les forces musculaires qui agissent sur le vagin ont une contribution substantielle dans le processus de miction.

Après avoir brièvement décrit l’anatomie et les propriétés de l’urètre, puis la théorie récente du chirurgien P.Petros sur l’ouverture de l’urètre chez la femme, nous présentons notre modèle en trois dimensions. Le modèle tient compte de la structure et de la composition complexe du tube, de la pression interne provoquée par le fluide, des contraintes et forces d’ouverture externes. Le mécanisme d’ouverture de l’urètre féminin pendant la miction apparaît comme une combinaison complexe entre la pression interne générée par la contraction de la vessie et de forces extérieures. On s’intéresse particulièrement aux résultats en trois dimensions obtenus en fonction des paramètres de pression interne, de structure du tube, de propriété des matériaux, de contraintes et de forces externes.

On montre que la simulation réaliste des dimensions et de la forme du tube ouvert n’est possible que lorsque tous les facteurs ci-dessus sont pris en compte. On observe finalement que le modèle est capable de simuler efficacement la géométrie complexe du tube de l’urètre observe expérimentalement aux rayons X.

Le modèle pourra être utilisé dans un futur proche pour étudier les effets des propriétés des différents tissus de l’urètre et les causes des dysfonctionnement lors de la miction.
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4. The Finite Element Analysis of the Urethra

In this chapter the finite element model of the urethra will be introduced. The geometry, material properties and assumptions and simplifications will be presented. Furthermore, the idealization of the boundary conditions will be described. Different configurations of the applied loads and boundary conditions were evaluated. All of the analyses were carried out with ABAQUS.

4.1. Urethral Histology and Anatomy

The urethral mucosa is surrounded by a relatively thick layer of submucosa composed primarily of longitudinally arranged collagen fibres. A rich vascular plexus is enclosed in the submucosa all along the urethral length. Outside the submucosa an inner, longitudinally arranged layer of smooth muscle exist, which again is surrounded by an outer thin, circularly arranged layer of smooth muscles (Fig. 4.1).

The thickness of the outer circular layer is almost the same throughout the entire length of the urethra, although it may be slightly thicker in the proximal third. The relatively thick striated muscle encircles the inner urethral layers with its thickest part at the middle two-thirds of the urethra. The muscle is horseshoe-shaped at the proximal and distal parts of the urethra [3]. The urethra is not entirely straight but has slight bends and follows an S-shaped course [9].
4.2. Finite Element Model Geometry

The geometry was taken from various sections through the urethra. In general, the structure of the urethra of different persons may differ. The structure is very complex and not symmetric. It changes longitudinally (Fig. 4.2, 4.3 and 4.4). Therefore it is obvious that the geometry of the structure for the model has to be simplified. Some features of the structures have been neglected, the dimensions were generalized. That was done to obtain a model with a reasonable fidelity of the real structure, by keeping the model geometry as simple as possible.

Till now, there is still no satisfactorily way to determine the cross sectional shape of the urethra during voiding. Data that is available is mostly derived from post-mortem studies, and therefore do not reflect the physiological form. The current assumptions regarding the cross sectional shape of the urethra range from slit-like to circular [9].

In following analyses, the cross sectional shape of the urethra will be assumed to be circular and in the longitudinal direction assumed to be a tube with a knee almost at the middle of the tube. (Fig. 4.6)
The cross sectional geometry of the finite element model is shown in Fig. 4.5. The several types of tissues that can be found in the actual urethra are reduced to three major types of muscle tissue. Material test data was available for these tissue types.

Fig. 4.3 Urethra cross section [17]

Fig. 4.5 Geometry of the urethral cross section for the finite element model
For the model and analyses, the following assumptions were made:
- the structure and the dimensions of the cross section do not change longitudinally
- the models’ cross section is based on the structure found in the mid-urethra

The length of the urethra was assumed to be 37.5 mm [5],[6], the diameter of the various muscle coat were estimated by examining several cross sectional pictures of the urethra, like Fig. 4.3. There is a knee at the middle of the urethra that fits the curve that appears in the x-rays (Fig 4.6). Another assumption made for the finite element model was to neglect the submucosa.

The inside of the urethra, the submucosa (Fig. 4.1, 4.2 and 4.4), is made of a porous structure, a tissue with a rich vascular system. These vessels are formed in such a way that the flow of blood into large venules can be controlled to inflate or deflate them. The function of this inner coat is to help to assure the waterproof state of the closed urethra.

Since the submucosa (painted in grey in Fig. 4.7) can be assumed to be a very soft material [14], the energy to deform it can be neglected, compared to the muscular coat of the urethra. During micturition the submucosa is being compressed to a thin layer [4], and the inner applied pressure to urethra is at whole transmitted to the muscular coat. Thus in the following analyses the submucosa is not modelled.

Fig. 4.6 X-rays of the urethra, at rest.

Fig. 4.7 Schematisation of the urethral muscular coat’s geometry during continence and micturition phase [4]
4.3. Material Properties of the Finite Element Model

The structure of the urethra in the model comprises different materials. These are smooth muscles (longitudinal and circular), striated muscle, and bladder smooth muscle (also called trigone).

Since only test data from uniaxial tensile tests were available, isotropic behaviour is assumed. This is a very rough simplification of the real properties of the tissues. In reality, the muscle fibres are orientated, so there is a significantly different behaviour in the various directions.

Furthermore it is assumed that the tissues consist solely of muscle fibres, neglecting that muscles are interspersed with connective tissue, which is especially true for smooth muscle. This connective tissue is comparable to the matrix material in composite materials.

The material is assumed to be fully incompressible, due to the high water content in human tissues.

Test data for the various muscle materials represents the behaviour in the resting state. Muscle activation changes the mechanical properties of the material considerably. In reality, during micturition the muscles of the urethra are assumed to be relaxed. According to this, since taken from specimens in the passive state, the material test data seems to be appropriate to reflect the actual mechanical behaviour of the muscles in the urethral wall.

This is a static analysis, thus time dependent material behaviour such as viscoelasticity, creep and stress-relaxation is not modelled.

Mechanical behaviour in vivo and in vitro differs. The test data used for the material model was obtained in vitro. Therefore the behaviour of the modelled structure is not representing the living organ.

Residual stresses (and strains) are inherent in many biologic tissues. There is no data for the urethra; however, it is likely that the urethra is no exception. It is supposed that such a prestressed wall tissue helps to close the urethra in the resting state. Residual stresses are not modelled.

A summary of the assumptions regarding the material behaviour:
- Muscles consist solely of muscle fibres.
- Muscle activation is not modelled => passive mechanical response for muscle material.
- Fully incompressible.
- No time dependence (no viscoelasticity, creep, etc.).
- Due to in vitro material test data: only in vitro behaviour.
- Isotropic (so no distinction can be made between circular and longitudinal muscle).
- Prestress (residual stress) in the materials is neglected.

Constitutive model - Hyperelastic material: polynomial model for strain energy potential
In order to create the materials of soft tissues, Joerg had to find the Polynomial function of the strain energy function that fits best the real curve found in other books.

Material test data was extracted from stress strain curves found in [12] and [15]. Values for nominal stresses and strains from uniaxial tensile tests were taken from curves (Fig. 4.8) printed in these books and transformed into appropriate dimensions. Unfortunately, only very few data exists describing the mechanical properties of soft tissue.

Since the material data was gathered by third persons, and the quality of that data is suboptimal (due to the data extraction by measuring values from a depicted curve), a certain error is involved in the analyses.

The stress and corresponding strain values were given to the finite element software. ABAQUS is then able to calculate coefficients for the chosen strain energy function, based on these material test data. There are several polynomial models available. The coefficients are written in an output file. Then a simple one-element test was done, to check how the models fit the test data. A second order polynomial model fits the test data best. The ABAQUS material check claims the material model to be unstable at certain strains. These strain values are written in the output file.

As seen later, when the reduced polynomial was used, the structure was unrealistic softer than the model with a full polynomial strain energy function. Therefore a polynomial of $2^{nd}$ order was used. But there is still the problem that the material model may become unstable at higher strains. Providing test data for additional deformation modes would help to increase the stability, but unfortunately, only uniaxial test data was available. Using additional data points by interpolation of the given data alleviates the stability problem as well. The best results regarding the stability, i.e. the highest critic strains were achieved when the test data curves were approximated by an exponential function, although one has to sacrifice fit accuracy. Values from that function were then used to calculate the constants for the strain energy function.

<table>
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<th>Maximum isometric tension $P_0$ ($10^2$ N)</th>
<th>Cross-sectional area (mm$^2$)</th>
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<tr>
<td>a, whole muscle</td>
<td>410</td>
</tr>
<tr>
<td>b, two-bundle preparation</td>
<td>42</td>
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<tr>
<td>c, two-bundle preparation</td>
<td>74</td>
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<td>d, single fiber</td>
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Fig. 4.8 stress-strain curve for smooth muscle (left), for striated muscle (middle), for the trigone (right) [12]
4.4. **Internal Pressure Load**

During micturition, the urethra is exposed to a fluid pressure load due to the contracting bladder. The maximum pressure value is found at the transition urethra/bladder and at the outlet the pressure is zero. In the following finite element analyses, the pressure load due to the fluid flow is assumed to drop linearly from the bladder to the outlet in a first model. We have then used a more realistic model taking into account the friction of the urethral tube on the fluid, describing the flow as turbulent and taking into account the change of the flow canal due to the deformable hyperelastic walls. Still additional iterations reinjecting the pressure calculated from the first simulation in the undeformed model again and so on in order to converge to the real internal pressure hasn’t been done in this paper. A more accurate model of the internal pressure could take this idea into account.

Measurements of the urethral pressure profile show that during micturition, an average bladder pressure of 60 cmH2O at the midurethra is needed to open the urethra [3], [4]. In [3], the required bladder pressure to open the urethra was determined by performing an inflation of a balloon catheter. This balloon was positioned at various segments of the urethra. Pressures were measured at cross sectional areas of the urethra, where the area was determined from the known dimensions of the balloon.

![Fig. 4.9 Relation of pressure response P, dilation CA, and the rate of dilation Q. The minimum pressure P₀ to open the urethra (CA=0) was determined by extrapolation. [3]](image)

According to this measurements, a pressure load of 60 cmH2O or 0.0058 N/mm² (the conversion is 1 N/m² = 0.0102 cmH2O) was assumed to represent the bladder pressure during micturition.

4.5. **Elements**

An incompressible material response cannot be modelled with regular elements (except in the case of plane stress) because the pressure stress in the element is indeterminate. If the material is incompressible, its volume cannot change under this loading. Therefore, the pressure stress cannot be computed from the displacements of the nodes; and, thus, a pure displacement formulation is inadequate for any element with incompressible material
behaviour. Hybrid elements include an additional degree of freedom that determines the pressure stress in the element directly. The nodal displacements are used only to calculate the deviatoric (shear) strains and stresses [8].

In the explicit analyses, it cannot be assumed that the material is fully incompressible. Since in ABAQUS/Explicit hybrid elements are not used, some compressibility had to be provided. A default value of 0.475 for the Poisson’s ratio was used. This makes the material somewhat softer than the actual material, but since the material is not highly confined, the results were regarded to be sufficiently accurate.

4.6. The approach

Already in the early models, done by J. Panzer, it became apparent that with a load value taken from the measurements [3], and the experimental material data no significant deformation occurred. This is a known problem, originating from too stiff material test data, as proposed in [4]. Therefore higher pressure values were used, to get reasonable deformations.

According to the Integral Theory, urethral opening and shape during micturition is mainly caused by the interaction of muscle forces acting on the vagina (on which the urethra is attached) and the suspending ligaments. To confirm this, a model reflecting these muscle forces and these suspensions was created, as well as a model with the combination of a pressure and a more discrete force load. The resulting shapes of the urethra were then compared to those seen in x-rays.

4.7. The models

For all models a consistent set of units was used (SI mm):
- Length in millimetres,
- Force in Newtons,
- Mass in tons,
- Time in seconds.

The results presented later are all given in these units.
Due to the symmetry of the problem, only one half of the tube was modelled. Therefore the nodes on the symmetry axis were constrained to no translation in the x-direction. Material properties were assigned as shown (Fig. 4.10).

To evaluate the behaviour in three dimensions a 3D model was created. It is obvious that there is the need to simplify the geometry, since the actual form of the urethra is far from a pipe-like shape. In addition to this the geometry and material properties vary for different individuals, alter with the age of the subjects etc. The geometry of the cross section and the material configuration was assumed to be the same along the length of the urethra.

**Loading and boundary conditions**

Actual boundary conditions (spatial constraints and loads) of the urethra in the abdomen during micturition (Fig. 4.11):

The movement of the distal half of the urethra is relatively confined due to firm connections to the pubic bone by connective tissue
and ligaments (Fig. 4.11). The remaining section of the urethra is free to move. At the transition bladder/urethra the bladder is connected to a ligament, which restricts motion vertically.

### 4.7.1. The first model

The first boundary conditions in the finite elements model that we have used, before the personal advice of Professor P. Petros was:

1. The symmetry of the tube was modelled by constraining the nods on the Y-Z cutting plane not to move along the X direction. This enables us to work only with a half cylinder representation of the urethra.

2. The fixation of the distal half of the urethra with connective tissue was modelled by constraining the nods on the X’-Z’ plane to move vertically along the Y’ axis, so that dilation is still possible.

3. Motion in the longitudinal Z’ direction was set to zero on the force at the distal end of the tube.

4. The insertion of the ligament on the bladder outlet of the urethra was modelled by fixing the nodes on a small area (approximately 1.5 x 2.5mm) on top of the proximal urethral tube in the three directions.

#### Loads

5. The pressure load representing the pressure of the fluid on the inner surface of the tube during micturition was modelled by applying a pressure load on small areas (4.5 x 2.5mm).

6. On the proximal end, the resulting force due to muscle contraction was modelled by applying a body force on a small element of the trigone (1.5 x 2.5 x 1.5mm) at the bottom of the tube, both along the vertical axis Y and the longitudinal axis Z.
4.7.2. The second model

The second model was then done in close cooperation with Professor Petros. Modifications were made to have a more realistic model, according to the surgeon’s experience.

**Boundary conditions**

1. The symmetry of the tube was modelled by constraining the nods on the Y-Z cutting plane not to move along the X direction. This enables us to work only with a half cylinder representation of the urethra.

2. The fixation of the distal half of the urethra with connective tissue was modelled by constraining the nods on the X’-Z’ plane in all directions, and by constraining the nods on the Y’-Z’ plane, the distal half only, in all directions also.

3. Motion in the longitudinal Z’ direction was set to zero on the force at the distal end of the tube.

4. The insertion of the ligament on the bladder outlet of the urethra was modelled by fixing a spring to a small area (approximately 1.5 x 2.5mm) of rigid skin on top of the proximal urethral tube.

**Loads**

5. The pressure load representing the pressure of the fluid on the inner surface of the tube during micturition was modelled by applying a pressure load on small areas (4.5 x 2.5mm).

6. On the proximal half tube, the resulting force due to muscle contraction was modelled by applying body forces on small elements of the trigone (1.5 x 2.5 x 1.5mm) on the bottom of the tube.
4.7.3. ABAQUS Explicit

Load

ABAQUS/Explicit uses explicit time integration to solve nonlinear dynamic problems. The process is static; however, ABAQUS/Explicit yields an efficient quasi-static solution if the time period of the event and the loading rates are chosen suitably.

To avoid sudden, jerky movements, and therefore exclude inertial effects, the pressure load was defined in a smooth way, using an amplitude definition for the load value like seen in Fig. 4.14.

![Amplitude vs. time graph](image)

Fig. 4.14 Load definition

Hourglassing

In the explicit analysis, reduced-integration elements were used. Linear reduced-integration elements tend to be too flexible because they suffer from their own numerical problem called hourglassing. A single reduced-integration element considered modelling a small piece of material subjected to pure bending (see Fig. 5.15).

![Deformation of a linear element with reduced integration](image)

Fig. 5.15 Deformation of a linear element with reduced integration subjected to bending moment [19]

Neither of the dotted visualization lines has changed in length, and the angle between them is also unchanged, which means that all components of stress at the element's single integration point are zero. This bending mode of deformation is thus a zero-energy mode because no strain energy is generated by this element distortion. The element is unable to resist this type of deformation since it has no stiffness in this mode. In coarse meshes this zero-energy mode can propagate through the mesh, producing meaningless results [19].

In ABAQUS a small amount of artificial "hourglass stiffness" is introduced in reduced-integration elements to limit the propagation of hourglass modes. This stiffness is more effective at limiting the hourglass modes when more elements are used in the model, which means that
linear reduced-integration elements can give acceptable results as long as a reasonably fine mesh is used [19].

Since the stable time step is determined by the smallest element length in the model, a coarse mesh increases the computation time. However, this also increases the effects of hourglassing, as it was seen in later models. The effects of hourglassing could be alleviated by a finer mesh, although the computation time vastly increased. However, the final deformed shape was approximately equal with both meshes, and since the stresses were of minor interest in the analyses, a relatively coarse mesh seems to be appropriate.

5. Results

Here the results of the explicit analyses will be presented. Different configurations of loads and boundary conditions were evaluated. Finally the results will be compared to the actual shape of the urethra during micturition.

5.1. The Models

According to Peter Petros’ Integral Theory, the “funnelled” shape of the urethra during micturition is mainly caused by muscle forces transmitted to the urethra. In the following chapter the results of various models simulating the loads on the urethra will be presented.

The magnitude chosen for the loads was somewhat arbitrary, since with the bladder pressure load of 0.0058 N/mm², as proposed in [3], no significant deformation occurred. Furthermore, no information was available regarding data describing the muscle forces pulling the urethra downwards. Referring to [4], where it is stated that experimental mechanical data, like the one that was used here, is too stiff est. factor of 100 (see discussion 6.5), the bladder pressure load value was increased by a factor of 100 to 0.58 N/mm². The same order of magnitude (applied to a volume) was used for the body forces, modelling the additional muscle force pulling the tube downwards.
5.1.1. Model with simulated fluid pressure

Linear pressure

In this model, the pressure of the fluid varies linearly; this was the first approximation of the conditions of the flow in the urethra during micturition.

Fig. 5.1 Deformation under linear varying pressure on inner surface of the tube

Fig. 5.2 Deformed state of the tube under a linear profile of pressure


Viscous Pressure

As explained in 2.4., we have calculated a more complex and more realistic pressure, its profile is shown on fig. 5.4. When applied to the urethra, without any external forces, the results are shown in fig. 5.5.
Fig. 5.5 Deformed state of the tube under a more complex profile of pressure

Fig. 5.6 Cross section of the deformed tube under a more complex profile of pressure
5.1.2. **Model with simulated forces**

**First BC**

In this model, the resulting muscle force pulling the urethra open during micturition was simulated by applying a body force on an area as shown in figure 4.12. The load and the prescribed displacement constraints are shown in Fig. 5.7.

![Fig. 5.7 Applied loads and displacement constraints](image1)

![Fig. 5.8 Deformed state of the tube under the applied loads](image2)

As can be seen in Fig. 5.8, only a very localized deformation occurred, and due to the type of loads, the cross section takes a very slit-like form, see in Fig. 5.9.
Fig. 5.9 Cross section of the deformed tube under the applied loads
Second BC

In this model the volume on which the body forces are applied is increased to half the length of the trigone. (Fig 4.13). The loads and the prescribed displacement constraints are shown in Fig. 5.10.

Fig. 5.10 Loads and prescribed displacement constraints

Fig. 5.11 Deformed state of the tube under the applied loads
5.1.3. Model with simulated fluid pressure (linear) and muscle forces

First BC

Then, a model combining a linear fluid pressure and an external muscle force was evaluated. The bladder pressure was stepped down to zero as previously, the muscle force was modelled with a body force applied on a small volume at the proximal end (Fig. 5.13).
Second BC

Then, a model combining a linear fluid pressure and an external muscle force was evaluated. The muscle force was modelled with a body force applied on a large volume of the trigone (Fig. 5.16).
Fig. 5.17 Deformed state of the tube under the combined load

Fig. 5.18 Cross section of deformed tube under combined load
5.1.4. Model with simulated fluid pressure (friction) and muscle forces

First BC

In this model, the urethra is submitted to an internal load (pressure of the fluid taking friction into account) and to external loads, applied only at the distal end of the urethra, as shown in Fig. 5.19.

Fig. 5.19 Model with fluid pressure load and muscle force simulated

Fig. 5.20 Deformed state of the tube under the combined loads
Second BC

Finally, a model combining a more complex fluid pressure and muscle forces was evaluated. The muscle force was modelled as a body force applied on a large volume of the trigone (Fig. 5.22).

Fig. 5.21 Cross section of deformed tube under combined loads

Fig. 5.22 Model with fluid pressure load and muscle force simulated
Fig. 5.23 Deformed state of the tube under the combined loads

Fig. 5.24 Cross section of deformed tube under combined loads
5.2. Time step

The total time step between the beginning of the ABAQUS job and its end is one second. Still, ABAQUS outputs its calculus every 0.05 seconds. The total load amplitude is reached at 0.25. As our model was a 3D model and quiet complex, the time needed to complete the whole job and reaching 1 second was too long (20 days). Still, as the ABAQUS/explicit gives a quasi-static solution, the system gives the right final shape at 0.25 in our case. In a 3 days time, our jobs were able to go as far as step 0.15 and it was sufficient enough in order to have a shape close to the final shape as shown below. As a result, most of our simulation are considered at a 0.15 step time in order to make relevant comparisons between them.

To give an example, we can look at three time steps for the 0906-bodyforces.odb, modelling the second model with a linear pressure (Fig. 5. 25). The urethra has evolved between 0.15 and 0.20 but not anymore after. The evolution on this model gives a good idea of the final shape at 0.15.

Fig. 5.25 Three time steps of the same model
5.3. 

One can see on figure 5.17, and even more on Fig. 5.26 hourglassing patterns. To minimize the hourglassing that we observed in some results, we have tried to do the same model but with a refined mesh. Still we had to abort this job because the step time was still under 0.05 after one week. As we didn’t have enough computers for Abaqus jobs we had to use a coarser mesh. Still, as we are more interested by the shape of our results, the results of the coarse mesh model can be accepted as an approximate solution, accompanied by a much lesser computation effort.

Fig. 5.26 Hourglassing Pattern
6. Discussion

6.1. Verification of the integral theory

Fig. 6.1 Only fluid pressure

Fig. 6.2 Only muscle force

Fig. 6.3 Combination of fluid pressure and muscle force
The usual theory assumed that the urethra opens uniquely because of the pressure of the bladder. But with our model, with only a fluid pressure load, the inner diameter doesn’t change enough: it almost corresponds to the opening seen in the X-rays, but the characteristic funnelled shape can not be seen with these results (Fig. 6.1).

An explanation comprises two effects. Firstly, the overall inflation of the tube cannot by reached, because the structure is to stiff (see 6.5).

However, this deformed state does not show the remarkable funnelled shape of the proximal section of the urethra, as seen in reality. So secondly, there must be other forces opening the urethra entirely and creating the funnelled shape. When applied in combination with the fluid pressure load, an additional load, representing the downward and longitudinal muscle forces, gave a good reproduction of the funnelled shape as seen in reality. This is a confirmation of the Integral Theory, which involves the participation of muscle forces during micturition.

A comparison of the different deformation states is given in Fig. 6.1 – Fig. 6.3.

6.2. Influence of pressure

Our first task was to calculate a more realistic pressure than the linear one, taking into account the friction with the urethral wall. The realistic pressure we have calculated participates more to the deformation of the urethra than the linear pressure. The opening of the urethra with the same boundary conditions is more important in length and in width with the pressure taking friction into account than with the linear pressure (as we can see in Fig 6.4-6 and Fig 6.5-7).

The results with the linear pressure are nearer to the shape of the X-rays photographs we are using than the results with the pressure taking the friction into account. One explanation of this fact is that we compare only with one x-ray photography. And we know that the shape of the urethra is very different from a woman to another. So we didn’t focus on this comparison, but on the importance of a future model as realistic as possible. The new pressure is indeed a step forward towards a more accurate model of the urethral opening as the values calculated are closer to the ones found experimentally.
6.3. **Influence of Boundary conditions**

Our major work has consisted in improving the boundary conditions of the previous model.

The boundary conditions representing the fixation of the connective tissue was in the first place just a fixation of a third of the distal end of the urethra, and then it evolved to a half, to finally be modelled by a complete fixation of the distal half of the urethra. We can see the influence of these different models on Fig. 6.4-6.5 and Fig. 6.6-6.7. With the advice of Professor Petros, we have considerably restrained the distal half of the urethra.

The boundary condition representing the proximal ligament has also evolved. In the first place, it was fixed to a surface of the tube and this surface was fixed in all the directions. Our first idea was to insert a spring that would represent a more realistic behaviour of the ligament. However in our first tries the spring transmitted force was only applied to a point on the surface (the results weren’t satisfactory as it resulted in an infinite force on a single point). So we added a rigid skin on the same surface as in the first model and we fixed the ligament on the rigid skin, in order not to apply an infinite punctual load. We can see that these new boundary conditions improve the funnelled shape of the urethra model (Fig. 6.4-6.5 and Fig. 6.6-6.7).
The boundary conditions representing the external muscles forces were in our first models pressure forces acting on the outside surface of the urethra. But, as the pressure is always perpendicular to the surface on which it is applied, the model was really instable, especially when we added the spring. So we tried several other ways to apply a force (not punctual) in permanent directions. Finally we have applied body forces only on the stiffer muscle of the urethra, the trigone. In our first model based on various bibliographical sources, before meeting with Professor Petros, we applied these forces only on the proximal end of the urethra. After the meeting and a closer interpretive look at some X-ray photographs, we applied these forces on the proximal half of the trigone.

6.4. Cross Section Shape

The initial cross section shape of the urethra is a circle. After applying the loads, the cross section shape of the urethra is elliptic. The longer axe of the ellipse is the vertical and the shorter the horizontal (as seen in Fig. 6.8, right).

Fig. 6.8. Urethral cross sections during micturition [9] (on the left), and shape of our model cross section (on the right).

This elliptical force is very predictable with the Integral Theory because of the loads; if there was only the pressure load, the shape of the cross section would be more or less circular. The presence of the external muscle forces pulling downward (among others) is giving the elliptical shape.

In [9], X-ray pictures using densitometry of the urethra during micturition were taken (see Fig. 6.8 on the left). These shapes are more complex than our cross section shape; this is caused by the complex form of the urethra when at rest (form that we modelled as a circle). But we can see that the global cross section shape is elliptical, as in our results. There was no indication in the article to know which the directions of the axis were. We can assume that this x-rays confirm the Integral Theory as well (if rotated by 90°).
6.5. Urethral muscles

A big difficulty of our work has been to find a realistic order of magnitude in the definition of the internal pressure. The value of the internal pressure is directly linked to the order of magnitude used in the stiffness definition of the urethral muscles. The material data for the model is based on measurements done with dead tissues. A central discussion to this problem is the difference between living tissue and dead tissue. Indeed, biological muscle measures are made on dead tissue. We can expect a difference between living tissue stiffness and dead tissue stiffness as a living tissue has the ability to relax. The dead one in contrary has fixed fibres as the living processes are stopped and it cannot relax. Therefore the dead muscle should be much stiffer than the living one [4]. In that paper, a factor of 100 is found between the stiffness of a dead and a living tissue. This factor is found comparing the pressure needed to open the urethra experimentally (around $3 \times 10^3$ Pa) to the pressure needed to open a basic model of the urethra described as an isolated uniform longitudinal pipe (around $5 \times 10^2$ Pa). Still, even if the idea of a difference between the two stiffness is relevant, a factor of 100 doesn’t seem realistic according to the experimental view of surgeon P.Petros. In addition to this, their modelling of the urethra doesn’t take into account any external forces nor ligament that might have an influence on the pressure needed to open the urethra.

Still, in our work, a pressure 100 times bigger than in reality had to be taken (even if we are using muscle forces in our model) to open the urethral tube as seen on the X-Rays. With a more realistic pressure only 10 times bigger than the one measured experimentally, we didn’t get any relevant dilation of the tube as shown in Fig 6.9.

![Fig. 6.9. Model with a ten time smaller pressure than the one we used in our work](image)
7. Conclusion

Geometry

As there is no common theory on the anatomy of the urethra, the overall shape of our model was inspired by X-Ray photographs. The detailed composition of the urethra muscles is based on various sources available at the time of our work and on the surgeon’s point of view. Still, all these data differ in some way from subject to subject, and vary with age, surgery and pregnancy. Our final model is a simplification of the very complex reality, modelling only the muscles that have a real influence in the opening of the urethra, taking into account the cross and longitudinal changes of the material distribution. As we were trying to keep the model as simple as possible, we assumed the pulling forces from the bladder to have the order of magnitude of the opening pressure force.

Material properties

Information on the mechanical properties of the urethral muscles is even more limited. Consequently to a lack of data on the mechanical behaviour of the smooth and striated muscles of the urethra at the time of our work, animal data have been used. Maybe new material data on the urethral smooth and striated muscles will improve the accuracy of the simulation. Furthermore, the material was created in a process of idealization, neglecting time dependence and muscle anisotropy. Still, our model is prepared for the distinction between circular and longitudinal muscles as it is already taking it into account in the geometry modelling. For example, a more detailed model could take into account the anisotropic data of muscles.

Boundary conditions and loading

In our last model with a spring ligament and a proximal-half downward and longitudinal uniform force, the boundary conditions are in accordance with various sources and professor Petros’ advice. Still, a more accurate model of the bladder pulling force while contracted would be possible by modelling the whole bladder. A future model could take this point into account. Furthermore, the internal pressure forces used to open the urethra are two orders of magnitude higher than experimental data. This can be explained by the fact that the only material test data on animal muscles found for the striated and smooth muscles, and used in our model, result in a too stiff material. Maybe new mechanical tests on human urethral smooth and striated muscles can confirm this explanation and solve this problem.
8. Bibliography


http://ligwww.epfl.ch/~maurel/Thesis98.html


[8] ABAQUS/Standard Documentation


http://www.me.ic.ac.uk/materials/staff/tissue.pdf


[18] Joerg Panzer, A mechanical model of the female urethra, Diploma Thesis, University of
Western Australia, 2003

[19] ABAQUS/Explicit Documentation

[20] Werner Schäfer, analysis of active detrusor function during voiding with the bladder
working function, 1990