A computational analysis of the effect of supporting organs on predicted vesical pressure in stress urinary incontinence

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Abstract

Stress urinary incontinence (SUI) or urine leakage from urethra occurs due to an increase in abdominal pressure resulting from stress like a cough or jumping height. SUI is more frequent among post-menopausal women. In the absence of bladder contraction, vesical pressure exceeds urethral pressure leading to urine leakage. The main aim of this study is to utilize fluid-structure interaction techniques to model bladder and urethra computationally under an external pressure like sneezing. Both models have been developed with linear elastic properties for the bladder wall while the patient model has also been simulated utilizing the Mooney-Rivlin solid model. The results show a good agreement between the clinical data and the predicted values of the computational models, specifically the pressure at the center of the bladder. There is 1.3% difference between the predicted vesical pressure and the vesical pressure obtained from urodynamic tests. It can be concluded that the accuracy of the predicted pressure in the center of the bladder is significantly higher for the simulation assuming nonlinear material property (hyperelastic) for the bladder in comparison to the accuracy of the linear elastic model. The model is beneficial for exploring treatment solutions for SUI disorder.

Keywords Computational fluid dynamics • Urinary tract • Stress urinary incontinence • Finite element method • Fluid-structure interaction

1 Introduction

Diseases associated with urine and genital tract are costly conditions prevalent among both men and women [1]. The most critical subset of these problems is urinary incontinence [1]. Stress urinary incontinence (SUI) is the result of mechanical pressure and is very common among women. Approximately 10 million women worldwide and over 6.5 million in the United States suffer from SUI [2]. Despite the fact that SUI is a common issue for elder women and women who have given birth, recent research shows that SUI is an issue that cannot be overlooked in female nulliparous athletes with the incidence varying from 12.5 to 80%. However, SUI is not reported for young athletes because of the fright of embarrassment [3]. In this type of incontinence, any activity that increases abdominal pressure (including laughing, coughing, sneezing, and straining) leads to urine leakage as a result of urethral sphincter weakness. Indeed, an increase in abdominal pressure in the absence of bladder contraction raises the vesical pressure to a level that exceeds urethral pressure, leading to involuntary loss of urine which mainly characterizes SUI. With respect to the reports published in 2001, costs of urinary incontinence treatments in the United States exceeded 16.3 billion US dollars. The reports indicate that 75% of total costs were spent on diagnosis and treatment for women [1].

Although the main reason for SUI remains unknown, a large number of physicians believe that SUI is caused by injuries to the pelvic floor neuro-musculature, which mainly occur among those who have given birth vaginally [1]. While it is not a life-threatening condition, SUI can
detrimentially impact the quality of life. Up to now, biomechanical studies over female incontinence are dominated by three theories, including pressure transmission theory proposed by Enhorning [4], the integral theory proposed by Petrus and Ulmestan [5], and the hammock theory proposed by Delancy and Ashton-Miller [6]. In addition, all these three theories seem paradoxical and within themselves, there is no consistency about involved structures and tissues. On the other hand, analytical studies and organic structures’ modeling are a practical way to study biomechanical phenomena. Several computational studies have been carried out for comparing healthy and damaged supporting tissues to understand the pathophysiology of the SUI [7].

In addition to numerous urodynamic studies for SUI, there are emerging computational models in recent decades to investigate questions tightly related to stress urinary incontinence [6, 8]. Kim proposed a 2-dimensional model to study urethral closure during stress [9]. He proposed an axisymmetric model of the urethra and pelvic floor, in which a catheter was placed in the urethra. This model was developed to analyze active (muscle contraction) or passive (pressure transmission) contributions to urethral closure. The results revealed that an active contraction of the sphincter muscle and pelvic floor plays an important role in urine continence [9]. The findings showed that levator ani’s contraction and its connectivity to endopelvic fascia contribute to urine continence [9].

Haridas et al. developed a finite element model to study biomechanical characteristics of the inferior urinary tract and the pelvic floor [10]. Their primary investigations were related to the finite element modeling of the vagina, bladder, uterus, levator ani, endopelvic fascia, and rectum derived from a specific MRI. A good coherence was observed between the modeling results and the ultrasound images. In the latest studies, Spirka et al. developed a finite element model of stress urinary incontinence [11]. The most critical limitation of this model was employing a quasi-solid model for the fluid. To verify the model, they used limited urodynamic data.

In the current study, the complexity of the numerical model (specifically the geometry of the whole urinary system and the interactions of its components) was a primary challenge. It is more common to investigate a part of urinary tract since scrutinizing all parts is very intricate and time-consuming [12–14]. Here, the modeling is mainly focused on bladder and urethra. The preferred computational parameter is the fluid (urine) pressure due to its availability in urodynamic measurement. The bladder reaction to an external force (abdominal pressure resulting from a cough) is studied in both physiological and pathological conditions. To achieve this goal, two different computational models were considered to investigate urine dynamics in the bladder. Fluid-structure interactions (FSI) method is employed in the computational models. The required data for the assumption of the simulation are extracted from urodynamic measurements. Damaser and Lehman [15] determined that the material properties of the tissues are more important in predicting the mechanics of the bladder than the geometry of the urinary tract. Therefore, simplified models are considered in this study to reduce the computational costs without affecting the accuracy of the simulation results.

The current study works on extending previous studies to model the inferior urinary tract, as well as validating the model through comparison to clinical and urodynamic data. The preceding models were focused on one mechanism of urinary continence [10] or they modeled urine leakage [11]. This study is focused on the parameters contributing to leakage and compares physiological and pathological cases. Another contribution is employing FSI method to improve the accuracy of the modeling. Some previous models claimed to be FSI [11, 16], but in fact, the urine has been modeled as Lagrangian elastic bodies in these studies. By contrast, in this study, the urine flow is modeled using an equation of state for the Eulerian material. As a result, in comparison to previous contributions, the vesical pressure in the center of bladder has been predicted with a higher accuracy using a 2D axisymmetric model, in which the computational cost is significantly lower than full 3D models.

2 Methods

In this research, the stress urinary incontinence was studied using the data collected from urodynamic tests. Urodynamic tests are measurements done to examine the bladder’s function and sufficiency. The real tests differ from person to person. Some urodynamic tests are quite simple and can be performed in a physician’s office. Other tests necessitate expensive and sophisticated equipment to measure the quantity of pressure experienced by the bladder and urethra [17]. In the current study, the patient was instructed to come for testing with a full bladder. During patient urinate into a container, the amount of urine and the degree at which the bladder empties were measured. A thin, flexible catheter was then placed into the bladder through the urethra, and the volume of any urine remaining in the bladder was measured (post-void residual, or PVR). A minor scorching sensation may happen when the catheter is placed. The bladder was filled with water through the catheter until the patient had the first desire to urinate. The volume of water in the bladder was calculated at this point. Then, further water could be added while patient resists urinating until involuntary urination occurs. All of the experimental protocols followed were in accordance with the ethical standards of the institutional and national committee on human experimentation and were performed according to the international rules considering the human rights.

Collected clinical data (obtained using urodynamic test, as described above) were gathered from two cases: a 48-year-old continent female to construct the physiological model, and a
34-year-old incontinent female to build the pathological model. Deliberate sneezing during urodynamic measurements was a comparison criterion between clinical and computational data. In this research, cough and its impact on the inferior urinary tract in both physiological and pathological groups were studied. To achieve this goal, two computational models were developed: one without support structure as a pathological condition and the other with a support structure that prevents leakage from the bladder outlet resembling physiological groups. The support structure is a bowl-shaped diaphragm similar to the drawing in the anatomy textbooks [18, 19]. The external force was sneezing which occurs in 200 ms. Both models have been developed with linear elastic properties while the patient model has also been simulated utilizing the Mooney-Rivlin solid model. The measured parameter for verification was vesical pressure. The total time of the simulation was 200 ms. The initial external pressure (abdominal pressure) was zero at the initial state and remained zero for 10 ms. Then, the abdominal pressure increased to its maximum level (which corresponds to urodynamic measurements obtained by a rectal catheter) and gradually decreased to the initial level. This pressure was applied gradually to prevent a shock in the computational domain. The profile of the applied pressure is plotted in Fig. 1. During the process, the pressure in the center of the bladder (vesical pressure) changed with time.

To model both pathological and physiological conditions, two different geometries were considered. In the physiological model, pelvic diaphragm closes urethra and prevents urine leakage. In the pathological model, the support structure does not exist, leading to bladder dislocation which prevents urethral closure.

For geometry and boundary conditions, data were collected from the continent case for the physiological condition and the incontinence case for the pathological condition. For geometry construction, bladder capacity was considered. Abdominal pressure data was used for boundary conditions, and vesical pressure data was utilized for validation of numerical results. Furthermore, the sensitivity of the model to different parameters was considered to verify analytical data.

Inferior urinary tract geometry included the bladder, urethra, and the support structure. In the physiological model, support structure existed, while in the pathological model, it was dislocated and was upper than the bladder outlet (indeed, the bladder had a prolapse), so it had no impact on urine continence [6, 20–22]. Therefore, to reduce the computational cost, it was removed from the pathological model. Figure 2a and b show pathological and physiological models, respectively.

Geometry construction was performed based on the geometry of the bladder of the continent and incontinent females. By considering the fact that bladder geometry was presumed as a sphere, its diameter was set to 45 mm for both the pathological and physiological model, based on the vesical capacities mentioned in urodynamic documents. Sean et al. studied the urodynamic measurement of 42 women and concluded that average thickness of bladder wall with 200 cc would be 1.7 mm [23]. Similar findings presumed that the thickness of the bladder wall is approximately 0.9 mm in 56 cc [9, 15]. With respect to the previous findings [15], to simplify meshing process and fluid and solid net coherence, the bladder wall thickness was presumed to be 1.5 mm in both conditions.

The urethra was modeled as a cylindrical tube [24, 25]. In both models, the urethra was relatively closed, and fluid flow would increase its diameter. Given that the aim of this study was to calculate the vesical central pressure, the length of the urethra is not important. The model is axisymmetric, so the bladder outlet did not have a cruciate form, but fix support boundary condition was applied to the urethra orifice to prevent non-physiologic movement of the urethra to accurately model the bladder outlet during urination.

In the physiological model, the urethra passed through a support structure, and this structure supported the bladder. The support structure was modeled like a bowl-shaped based on its anatomy [19, 22]. Janda et al. reported that pelvic diaphragm supporting bladder is an oval-shaped with 140 mm of long-axis dimension and 122 mm of short-axis dimension [26]. To further simplify calculations, a circular bar was used with the radius taking the average of the two mentioned values, 131 mm and with 2 mm of thickness similar to levator ani’s thickness [27].

The Eulerian kinematic description is proposed for fluid flow. The fluid is modeled by an equation of state description, which describes the primary hydrodynamic response of the urine. In other words, the fluid was modeled as a quasi-solid material to make it easier to simulate the interaction of the Eulerian and Lagrangian regions [28]. As a result, no conventional fluid momentum description (such as Navier-Stokes equations) was employed for fluid flow simulation. A large enough fluid region is considered to encompass all Lagrangian details, in which the urine can flow and interact with the bladder. The regions of this Eulerian domain not filled by urine were considered as void, in which the urine can flow without the need of constructing a multiphase flow model. Given that each case was asked to sneeze deliberately.
during urination, the bladder was not completely full at the time of sneezing. Figure 3 shows the final computational model of the physiological bladder.

In a mathematical model, the boundary conditions are the key to the transformation from physical model to the computational one. Consequently, all the important aspects of the physical model should be considered by applying appropriate boundary conditions.

In the current research, four types of boundary conditions were used to construct both the physiological and pathological models. The most important boundary condition was the abdominal pressure, applied gradually as a distributed pressure normal to the surface of the top hemisphere of the bladder, which was correlated to the abdominal pressure during an actual human cough. The maximum pressure was derived from urodynamic measurements, so in both models, the maximum abdominal pressure was 50 cm H$_2$O (4.9 kPa). The profile of the applied pressure is plotted in Fig. 1. For the first 10 s, the pressure remains zero. Then, in the following 200 s, the pressure reaches its maximum level and decreases to zero again.

Another important boundary condition in mechanical simulations was the fixed support condition to prevent rigid body translation for the Lagrangian part. In the current study, the fixed support condition fixed the bladder in two different ways. In the physiological model, this condition completely constrained to the bottom rim of the support structure, which made this portion of the structure fix in space and prevented rigid body motion (no displacement was allowed in X and Y directions). Similarly, in the pathological model, this condition was applied to the bottom hemisphere of the bladder (due to the lack of support structure). In addition, to prevent the non-physiologic movement of the urethra, the fixed support boundary condition was applied to the wall of the urethra orifice to prevent movement along the X-axis (right to left). No fixity was applied for the Eulerian fluid, which means the urine was allowed to freely move around and interact with the bladder. Finally, the last applied condition was gravity, which was applied to both Eulerian and Lagrangian domains.

In order to discretize the model, a structured mesh was employed. The Lagrangian mesh including bladder, urethra, and support structure (exclusively for the physiological
model) was unstructured quadrilateral, but thanks to the sweep method, they were homogeneously dispersed in this region. The Eulerian part was also meshed with quadrilateral 4-node elements. The computational mesh is shown in Fig. 4, which shows how the Lagrangian mesh was placed on the Eulerian grid. The Lagrangian mesh description included 568 elements, and the Eulerian mesh contained 32,400 elements. Mesh sensitivity analysis was performed to investigate mesh independence of the numerical results (results are presented as supplementary materials).

In continuum mechanics, material properties dictate mechanical behavior. Recent findings [11] show that varying material properties have a minimal effect on the accuracy of the vesical pressure and displacements predicted by the model, so the current study presumed the bladder, the urethra, and the support structure as linearly elastic materials. Table 1 presents the constants of the linear equations. These tissues can also be considered as structures with isotropic hyperelastic materials [9, 29]. The mechanical property of a hyperelastic model is completely dependent on a scalar function of strain energy density which is considered as free energy in a unit of nondeformed volume of the material. The Mooney-Rivlin model is one of the most common models for hyperelastic materials [9]. Strain energy density function in the Mooney-Rivlin model is derived from Eq. (1) [29]:

\[ W = A(I_1 - 3) + B(I_2 - 3) \]  

in which \( A \) and \( B \) are experimental coefficient, and \( I_1 \) and \( I_2 \) are coefficients of first and second invariants of deformation tensor, respectively. In the patient model (the model that does not have the supporting organ), a simulation was also accomplished with the properties of Mooney-Rivlin model for the walls of the bladder so as to compare the results with the elastic model. The coefficient values in Eq. 1 for \( A \) and \( B \) are 7.5 kPa and 2.5 kPa, respectively [9]. The urine flow was modeled linearly at first, but there was no relationship with the clinical results. So, a polynomial equation was employed to describe the urine flow. Urine density is \( 1 \) g/cm\(^3\) [11]. In a linear model of urine, by neglecting thermal effects, shear stress modulus with the values of \( 2.2 \times 10^6 \) kPa is the only incorporated parameter [11]. In the polynomial equation of state of fluid flow, the pressure is derived from Eq. (2) [31]:

\[ P = A_1 \mu + A_2 \mu^2 + A_3 \mu^3 + (B_0 + B_1 \mu) \rho_0 e \]  

\( \mu \) stands for compression, \( \rho_0 \) is density at \( P_0 \), \( e \) is internal force per mass unit, and other constants are water constants which \( A_1, A_2, A_3, B_0, \) and \( B_1 \) are \( 2.2 \times 10^6 \) kPa, \( 9.54 \times 10^6 \) kPa, \( 1.45 \times 10^6 \) kPa, 0.28, and 0.28, respectively [31].

Abdominal pressure increases flow pressure, and consequently, urine exerts pressure on the walls of the bladder and the urethra. As a result, fluid-structure interaction is indispensable in computations. The shear stress at the interface leads to solid deformation or dislocation which impacts fluid flow [29–31]. To analyze this condition, the Arbitrary-Lagrangian-Eulerian equation was used, since it is simple and accurate. In this method, Euler equation was employed for fixed boundaries (fluids), Lagrange equation for moving boundaries (solids), and the Euler-Lagrange equation for the interactions [32]. Here, since sneezing is a quick dynamic incident, explicit method was employed.

ANSYS AUTODYN was used to analyze the model and solve its equations. The initial pressure in the Eulerian region was zero. To solve fluid and solid equations together, FSI and Arbitrary-Lagrangian-Eulerian equation were employed. The penalty method was employed after remeshing the deformed or dislocated region to save time. Therefore, the coupling of two Lagrangian and Eulerian methods automatically was started. Self-interaction was activated in Lagrange region.

![The computational mesh for physiological model (purple: sweep mesh of the bladder, green: sweep mesh of the support structure, blue: Eulerian mesh filled with urine, white: empty Eulerian mesh)](image-url)
and its tolerance was 0.2. The maximum time of solving was 200 ms, and the allowed tolerance of energy error was 4. The safety factor in time steps was set to 0.65. The strain rate in the Eulerian region was calculated based on the weight method, and Eulerian pressure was calculated based on the mean method. Also, mass scaling method was deactivated. The simulations were performed on a quad-core CPU with 6GB RAM.

3 Results

The comparison between pressure-time diagrams with respect to quantitative results is the most important part. Figure 5 shows the simulated vesical pressure changes for pathological and physiological conditions. As demonstrated in Fig. 5, the vesical pressure of the physiological model exceeds the vesical pressure of the pathological one, which is a direct result of the presence of the support structure to prevent the urine leakage. The same behavior is observed in collected urodynamic data, obtained from the continent and incontinent cases. The comparison is performed in Table 2.

Table 2 presents the computed maximum vesical pressure during sneezing and urodynamic measurements and compares the results. Similar to simulated vesical pressure and Fig. 5, the real vesical pressure in the continent case is higher than the measured vesical pressure of incontinent case. Table 2 also indicates that the results of modeling performed in the current study, including the incorporated material properties, geometries, and boundary conditions, have a difference of 1.3% with the vesical pressure obtained from the urodynamic tests.

Bladder wall dislocation contour is plotted to study vesical deformations and is compared to clinical data. Figure 6a and b represent the bladder wall deformation contour for the physiological and pathological model, respectively. These figures demonstrate the maximum deformation in the bladder wall, which are 4 cm and 2 cm in the continent and incontinent models, respectively. In addition to these contours, the overall progress of simulation (in case of bladder deformation), in both pathological and physiological models, are demonstrated in Figs. 7 and 8. According to the findings of this research, the simulation assuming hyperelastic material for the bladder (nonlinear model) has a higher accuracy in the pressure predicted in the center of the bladder in comparison to the simulation assuming linear elastic properties for the bladder.

4 Discussion

Although several studies have investigated bladder and urethra mechanical behavior, their results focused on bladder filling or emptying [15, 24, 29, 32-39]. Few studies focused on external forces and their impact on bladder condition. Zhang et al. published the most reliable results in the field of stress impact on the bladder [16]. That study modeled the organs involved during the jumping of height and presumed that urine leakage occurs as the urine enters the urethra. In their reported results, only the upper part of the urethra was filled with urine, while in the current study in the pathological condition, the support structure is missed and the urine fills the urethra completely which indicates that this criterion is not an appropriate criterion for urine leakage, even though urine leakage is not the preferred parameter in the present study. Similar to Zhang model [16], the current model developed did not simulate urine transmission through the urethra. Limitations in hardware resources and time-consuming processes lead to these defects. Although Zhang et al. studied two different stress incidents, our images of modeled bladder deformation match qualitatively with Zhang et al. results [16], which means that images of either research show that elements of the upper hemisphere are deformed and dislocated more than other elements. The least dislocation occurs in the inferior parts.

Kim et al. investigated the impact of a cough on a 2-dimensional model of the bladder and studied pressure
transmission theory and summed up abdominal pressure to bladder central pressure and bladder inner walls [9]. He also considered linear elastic properties, and his validation criterion was bladder and support structure deformation. In addition
to the similarities, the main features of the current study are the implementation of novel computational techniques and reasonable clinical relevance of the model assumptions, which differentiate it from similar studies. Furthermore, abdominal pressure is modeled as a dynamic impact (time-dependent pressure, which means the pressure is applied gradually) inserted in the superior part of the bladder, and bladder central pressure is calculated in the model. The abdominal pressure in Kim’s study is relatively equaled to clinically measured pressure for the pathological conditions in the current study. The maximum deformation of the current study is reported 2 cm (Fig. 6b) which is in line with the results of Kim [9].

With respect to the diagram shown in Fig. 5, bladder internal pressure in the physiological condition is more than both of pathological conditions, which matches with the urodynamic measurements by an error of 1.3%. The patient diagnosed with SUI has no control over urethral pressure, so any external stress to bladder leads to urine leakage and bladder central pressure reduction, while in physiological conditions, the pressure inside the bladder increases, since support structure prevents urine leakage. Similarly, Table 2 shows that the patient model with the nonlinear model for the bladder wall has more computational accuracy. Additionally, the bladder pressure value is closer to the real values and the same results quantitatively. Also, it verifies quantitative simulations in the current study. It was determined that the model could be validated through a comparison of the simulation results and the clinically measured data. The most important limitations of the current study are the computational cost associated with un-optimized meshing techniques, semi-real boundary conditions, and simplified geometries. Generally, overcoming each of these challenges would lead to a model with a higher level of clinical relevance, so for further studies, it is proposed that more recent FSI techniques should be applied and non-uniform meshing should be defined in the Eulerian region to increase accuracy. Boundary conditions should be defined in a way that does not limit the natural activity of the bladder and is more similar to the real situation. It is also proposed to utilize MRI images for geometry construction to improve the reliability of the results, in order to predict and differentiate pathological and physiological conditions.
5 Conclusion

The current study developed two models of stress urinary incontinence for physiological and pathological conditions to study urine dynamic flow in the bladder under abdominal pressure resulting from sneezing. The developed model and its validations showed that analytical methods and simplifying biological systems, like inferior urine tract, would successfully simulate the clinical results in a virtual environment. It shows that the investigations of biological systems do not necessarily require experimental sets. It illustrates that accuracy of the pressure predicted in the center of the bladder for the simulation having nonlinear material property (hyperelastic) is significantly more than the accuracy of the linear elastic model. The results of both patient and control computational models imply that the most deformation of the bladder under pressure occurs near the time that pressure reaches its maximum value. Although FSI method deploys complex and time-consuming calculations, it is very practical and accurate in modeling coupled systems including both solid and fluid phases. Results indicate that maximum deformation in bladder occurs when the vesical pressure is reaching its maximum level. The predictions made by the computational model indicate that numerical methods and simplified physics of biological systems such as inferior urinary tract are helpful to achieve the clinical relevant results. This model helps clinicians to understand pathological conditions of the lower urinary system, the mechanics of continence, the effects of bladder wall properties, and the anatomy of supporting organs in this regard. The current study could provide general insights into the mechanics by which the continence is maintained in women and the effects of bladder wall properties and the anatomy of supporting organs on the mechanics of continence during a rise in abdominal pressure. These insights are also beneficial for exploring treatment solutions for the problems caused by SUI. For further studies, it is proposed to utilize more recent FSI techniques, more accurate boundary conditions and geometry construction based on MRI images, in order to predict improved results in pathological and physiological conditions.

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