Proceedings of the World Congress on Engineering and Computer Science 2012 Vol II
WCECS 2012, October 24-26, 2012, San Francisco, USA

Computational Simulation of the Urinary System

Ghazaleh Hosseini[1], John .J.R. Williams[2], Eldad .J. Avital[3] , A. Munjiza[4] , Xu Don [5], James A. Green[6]

Abstract: **Recent advances in computational fluid dynamics (CFD), the availability of supercomputing power and the current understanding of the ureter make a realistic computational model of a ureter feasible. This study is aimed of developing a computational model of the ureter to examine urine flow. We use the in-house CFD code CgLes which is a highly parallelised three–dimensional fluid solver with second order accuracy in both time and space. It uses the Navier Stokes equations to simulate both laminar and turbulent incompressible flow of a fluid. The immersed boundary method has been implemented into CgLes to simulate moving boundaries and it will be used to simulate the interaction between the flow of urine and ureter. To simulate the wall of ureter, CgLes has been coupled with a world-class in-house Discrete Element Model (DEM) C code. An initial simulation has been carried out by applying a circumferential force to a flexible tube in two locations away from its inlet in order to simulate peristalsis motion. The tube was fixed at the inlet and outlet and we believe these deformations to be very similar to those of a real ureter. Further work will focus on the movement of urine down a ureter using the exact geometry from CT scans and the Mimics software.**

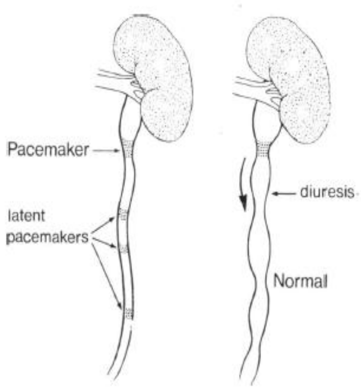
Key words**: (peristalsis motion, CFD, The immersed boundary, Navier Stoke equations)**

I. INTRODUCTION:

In the urinary system, peristalsis motion due to muscular contraction of the ureteral wall drives urine from the kidney to the bladder through the ureter as seen in figure (1b). The urine is expelled from the body through the urethra during voiding by increasing the pressure in the bladder. A primary pacemaker location is known at the beginning of the ureter which initiates the peristalsis motion. The reason of this activity can be the distension of the pelvi-ureteric region when urine is produced by the kidney. However, there are other parts of the ureter that are able to show the initiation of the peristalsis called latent pacemakers Figure (1a).

A common ureteral disease is Vesicoureteral reflux (VUR) which can be defined as a back flow of urine from the bladder into the ureters or kidneys due to a malfunction of the valve in the vesicoureteic junction.

[1] PhD student e-mail: g.hosseini@qmul.ac.uk ,[2] Professor, e-mail: j.j.r.williams@qmul.ac.uk ,[3] Reader, email: e.avital@qmul.ac.uk ,[4] Professor, email: a.munjiza@qmul.ac.uk ,[5]Post doctoral research fellow Email: d.xu@qmul.ac.uk, [6]Urologist consultant and director of research, email: James.Green@bartshealth.nhs.uk
[1,2,3,4,5]School of Engineering and material sciences, Queen Mary University of London, Mile End Road, London E1 4NS, United Kingdom, [6]Whipps Cross University Hospital, Whipps Cross Road, Leytonstone, London E11 1NR, United Kingdom



(a) (b)

Fig 1: a: the concept of primary and latent pacemakers in ureter, and b: behaviour of ureter during diuresis. [1]

Recent advances in CFD and the availability of supercomputers make a realistic computational model of a ureter and peristaltic flow possible and it can provide a tool to develop and explore further research. Extensive literature [2, 3, 4] has been devoted to finding mechanical properties of the ureter in both animals and humans and the majority of the studies indicated a viscoelastic behaviour for the ureter. The first attempts to create a model were based on the theoretical analysis and numerical solutions for viscosity dominated flow through a uniform tube with peristaltic motion [6]. In their study the hydrodynamics of flow through a distensible tube of a finite length with a peristaltic motion is discussed. Later, different rates of the frequency of peristaltic motion have been applied to the same model. The dynamics of the upper urinary tract have been investigated extensively and the pressure/flow relation in different conditions has been studied. [5, 6, 7, 8, 9, 10]. However, the fluid structure interaction has been not stated in any of the study mentioned above.

Experimental techniques which have also been used to quantitatively measure ureteral dynamics including; time resolved dynamic contrast-enhanced magnetic resonance urography for the evaluation of ureteral peristalsis using a data sharing 3D gradient echo sequence with spiral k-space filling [11], Color Doppler Sonography to evaluate ureteric jets in order to define patterns of flow and ranges of flow values in an asymptomatic population [12].

Vahidi and Fatouraee [13, 14] assumed a rigid surface with contact in solid wall that could simulate peristaltic motion and used the fluid structure interaction method to simulate the interaction between the flow of urine and the ureter. The results included the position of the highest shear

ISBN: 978-988-19252-4-4
ISSN: 2078-0958 (Print); ISSN: 2078-0966 (Online)

stress and possibility of reflux on the ureter wall. Also the effect of some factors such as ureteral wall compliance and different pressure amongst the kidney and bladder on the efficiency of the peristaltic process was investigated. Although they developed their model by introducing a nonlinear hyperplastic material model, (Arruda–Boyce model) and indicated a possibility of the urine reflux to the kidney at the beginning of the wave and it decreases by distally propagating wave through the ureter [15].

Application of CFD in order to simulate a realistic model of the ureter is a very new topic which can open many doors to future research in various aspects and this has motivated our group to work on this subject. The aim of this study is to develop to a computational simulation of the contraction of a ureter by using our in-house codes as an initial step towards further investigations of the reflux phenomenon and other ureteral abnormalities.

II. Methodology:

We used the in-house CFD code CgLes [16] which is a highly parallelised three–dimensional fluid solver with second order accuracy in both time and space. To simulate the deformability of the ureter, CgLes has been coupled with a world-class in-house Discrete Element Model (DEM) C code [17, 18]. The combined code is capable of modelling the movement of solids, including particles such as kidney stones in the ureter, and red blood cells.

i. Geometry of the Model:

We assumed a computational box of 4 blocks with the gird resolution of 32x32x128 for the fluid domain. The grid cell length in the y and z directions is about 0.037cm and 0.06 cm in x directions. For solid the length of tubes around 8 cm gives 10353 elements yielding and an element length of is 0.1 cm see figure (2) for illustration. One should note that we are interested in the flow inside the tube (ureter) and any small residual flow created in the computational domain outside of the tube due to its motion is of little interest for this study.

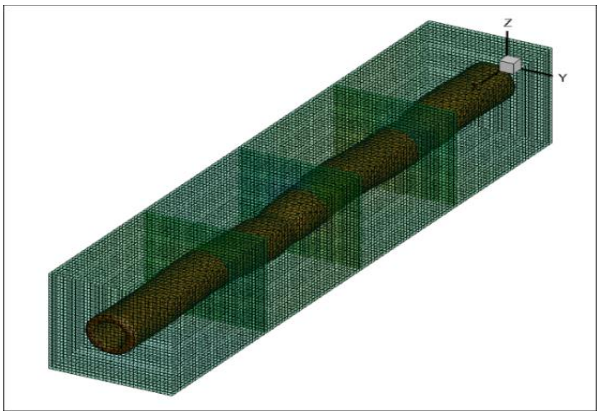


Fig 2. Geometry of current model.

ii. Contraction simulation:

To simulate the contraction, we defined two separated windows at the beginning and in the middle of the tube which could move down the ureter with a velocity of 2.4 cm/s which is similar to that in a human while applying a circumferential force to close the ureters cross sectional in order to push the urine down the ureter. The purpose of this design was to mimic the pacemaker activity in ureter to achieve a realistic contraction simulation.

iii. Fluid and solid model:

To simulate the walls of the ureter, we assume it is linear elastic homogeneous material with a Young's modules 5Kpas. No-slip boundary conditions on the tube walls were applied for the fluid flow. A mean pressure difference around 0.007 and 5 cmH_2O was applied as an inlet pressure boundary condition in the stream-wise direction. The outlet and inlet positions of the tube were been fixed and the fluid (Urine) is assumed to be laminar and Newtonian, viscous and incompressible.

iv. Fluid solid Interaction:

The immersed boundary method has been implemented into CgLes to simulate moving boundaries and was used to simulate the interaction between the flow of urine and ureter. The immersed boundary (IB) method is used to tackle such complex FSI problems in an easy and straightforward manner. The fluid motion equations are discretized on a fixed Cartesian grid. An extra singular body force is added into the momentum equation to take solid boundary into account (equation 1, 2):

$$u^{n+1} = u^n + \delta t\left(\frac{3}{2}h^n - \frac{1}{2}h^{n-1} - \frac{3}{2}p^n + \frac{1}{2}p^{n-1}\right) + f^{n+1/2}\delta t \tag{1}$$

$$f^{n+1/2}\delta t = D\left(V^{n+1} - I\left(u^n + \delta t\left(\frac{3}{2}h^n - \frac{1}{2}h^{n-1} - \frac{3}{2}\nabla p^n + \frac{1}{2}\nabla p^{n-1}\right)\right)\right) \tag{2}$$

v. Governing Equations:

The governing equations for the fluid are incompressible Navier Stokes (NSE) which is continuity, momentum equations

$$\rho\left(\frac{\partial \mathbf{v}}{\partial t} + \mathbf{v}\cdot\nabla\mathbf{v}\right) = -\nabla p + \mu\nabla^2\mathbf{v} + f \tag{3}$$

$$\nabla.\mathbf{v} = \mathbf{0} \tag{4}$$

Where $\mathbf{v}$ is the flow velocity, $\boldsymbol{\rho}$ is the fluid density; $\boldsymbol{\mu}$ is the fluid viscosity, $\boldsymbol{p}$ is the pressure and $\boldsymbol{f}$ is the body force. The solution can be obtained by using the finite volume method and applying the projection method to ensure continuity. The Navier Stoke equations are discretised by a second order Adam_Bashforth method in terms of time. Spatial derivatives, diffusion and convection terms are approximated using the second order finite difference method. A conjugate gradient method is mainly used to solve the pressure Poisson equation in CgLes codes.

III. Result and discussion:

We have to-date simulated laminar urine flow in a tube using published mechanical properties of ureters and we assumed that the fundamental reason for urine flow is mainly due to the contraction of the walls (pacemaker activity) and also the pressure difference between the kidney and bladder. We believe these deformations are more similar to those of a real ureter. See the figure (3) for the time sequence of wave propagation of the ureter under the action of application force.

We have simulated two different conditions, when contractions are applied in present and absence of pressure difference between the kidney and bladder.

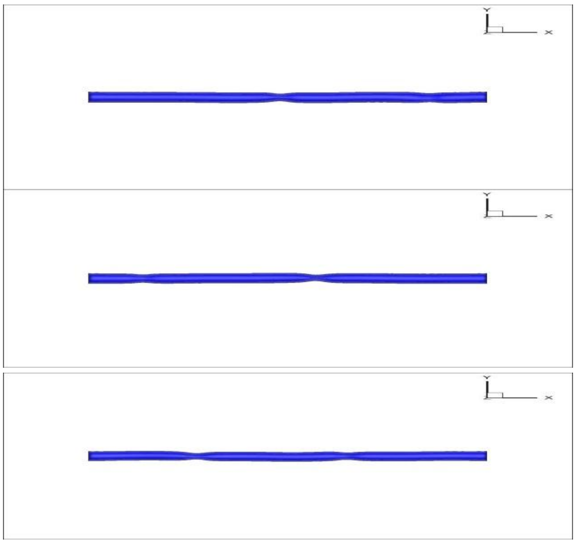


Fig 3. Simulated wave propagation in ureter.

Figures 4 and 5 show the pressure and velocity distribution along the ureter in two separated contractions simulating the action of the pacemakers along the ureter while no pressure difference over ureter was applied. Increase in pressure and velocity is observed around the contractions.

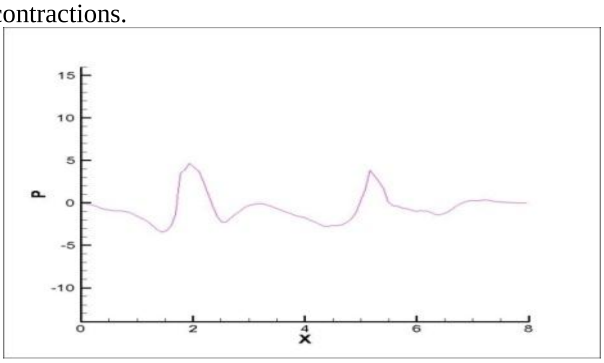


Fig 4. Instantaneous pressure distribution (dyn/cm2) along the centre line(cm).

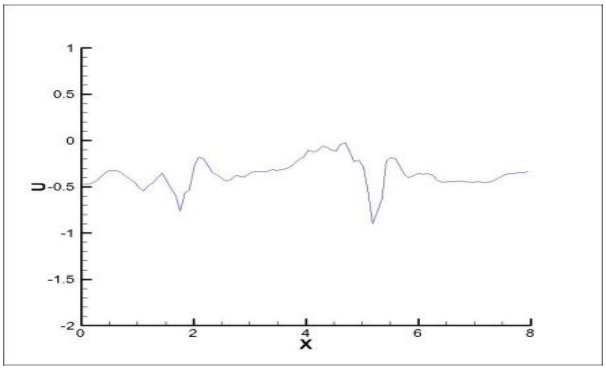


Fig 5. Instantaneous velocity distribution (cm/s) along the centre line.

For the other two simulations, two pressure differences of 0.007 and 5 cmH_2O were applied between the inlet and outlet of the flexible tube. It is worth mentioning that pressure difference between the bladder and the kidney can be between 0 to 5 cmH_2O and depends upon the hydration in normal and healthy ureter.

The pressure driven flow computation was validated by stiffening the tube's wall and comparing the velocity profile to the analytical solution of Poiseuille solution, achieving excellent agreement. Figure 6 and 7 shows an increase velocity in the contraction area in comparison with the situation without pressure difference

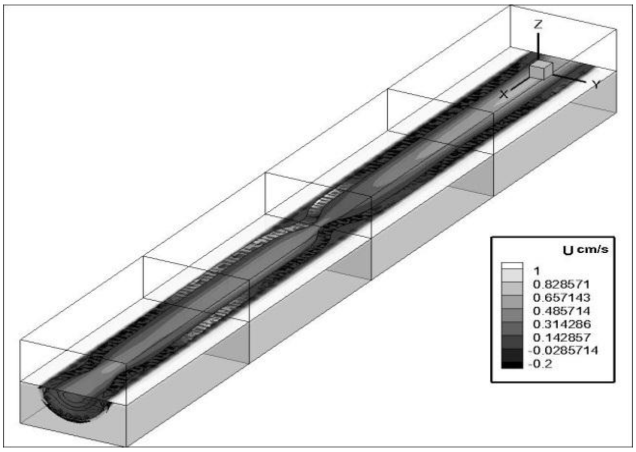


Fig 6. Instantaneous velocity contour in x direction when pressure difference is 0.007 cmH_2O.

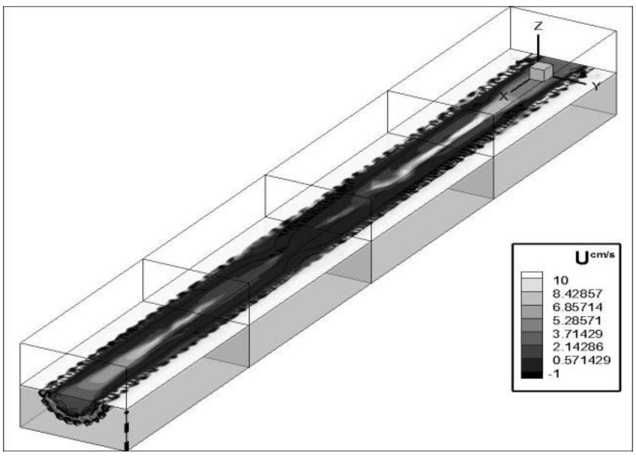
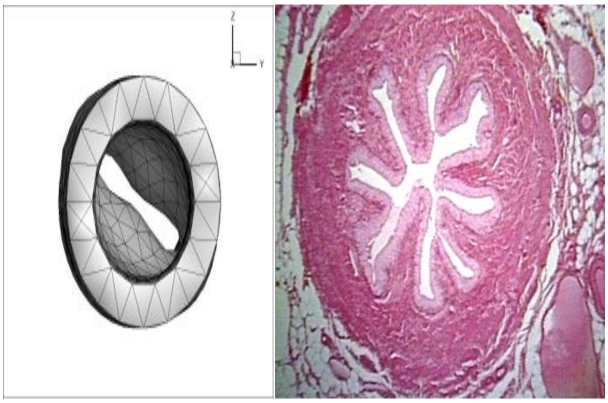


Fig 7. Instantaneous velocity contour in x direction when pressure difference is 5 cmH_2O.

ISBN: 978-988-19252-4-4
ISSN: 2078-0958 (Print); ISSN: 2078-0966 (Online)

The velocity contours in both the two pressure differences show an increase in the velocity magnitude in the middle of the tube as the pressure difference condition is applied. However, the velocity magnitude is somewhat too high for the high pressure difference of 5 cmH_2O as the flow rate of urine is known around 0.1 to 3 ml of urine per minute. This is likely due to the difference between the shape of cross sectional area of the ureter and our model. (Figure 8)



(a) (b)

Fig 8. (a): cross sectional area of the current model during the contraction, (b): cross sectional area of human ureter [19].

As seen from Figures 9 and 10 the lack of pressure difference causes a significant back flow at the beginning of the contraction, indicating a possibility of urine reflux. In comparison, the back flow is not seen when the pressure difference was applied, pointing to a smaller possibility of reflux.

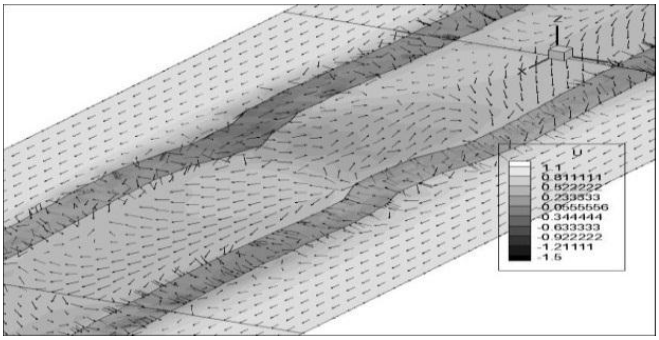


Fig 9. Instantaneous velocity vector in absence of pressure difference of 5 cmH_2O.

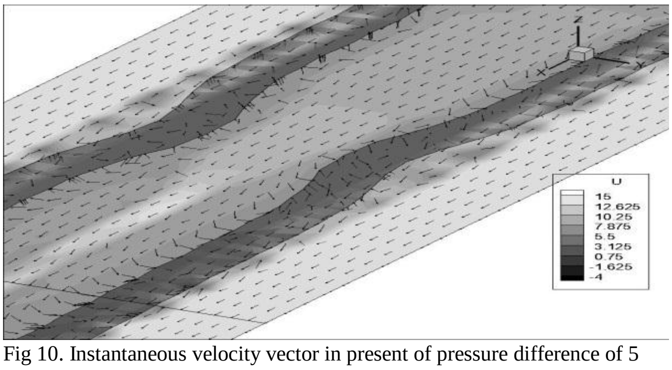


Fig 10. Instantaneous velocity vector in present of pressure difference of 5 cmH_2O .

IV. Conclusion and further work:

In this paper, a fluid-structure simulation technique was introduced to study two contractions in a small tube mimicking human ureter. The results showed an increase in pressure and velocity during the peristalsis around the contraction zone. A possibility of urine reflux was identified for a low pressure difference over the tube. Although the contraction design is very similar to the known physiological function, our model still needs improvement. We assumed the tube wall to be of an elastic homogenous material and used an initial circular shape for the tube's cross sectional area. The latter is crucial as it directly relates to the magnitude of the force needed to simulate the peristalsis and affects the flow field inside the ureter. We already modelled the exact geometry of the ureter's cross-sectional star shapes from CT scans and we are in the process of feeding this geometry to our in-house simulation software. Viscoelastic behaviour of the ureter has also been modelled into our codes and the results are expected to be reported soon.

REFERENCES:

[1] Richard G. Notley , The structural of basis for normal and abnormal ureteric motility ,Ann.Roy.coll.surg Engl.1971,vol.49.

[2] Apter, J., Mason, P. (1971) Dynamic Mechanical Properties of Mammalian Ureteral Muscle Technical Report, American Journal of Physiology: USA, pg. 266 - 271.

[3] F.C.P YIN, Y.C.FUNG, Mechanical properties of isolated mammalian ureter segments, American Journal of Physiology, Vol. 221, No. 5, November 1971.

[4] Nihon Hinyokika Gakkai Zasshi.. Mechanical property of the human ureter, Nakagawa Y journal, 1989 Oct;80(10):1481-8 also available in Helen E . Kambic , Biomaterial mechanical properties , page 275-282.

[5] . Karnak, I. Et al (2001) The effects of flow rate, length and external pressure upon the pressure required for fluid to flow through a ureter article, BJU international: Turkey, pg. 335 – 338.

[6] Griffiths, D. (1986) Dynamics of the upper urinary tract: I. Peristaltic flow through a distensible tube of limited length technical report, Erasmus University Rotterdam: Netherlands.

[7] Jiménez-Lozano, J., 2009, "Peristaltic Flow with Application to Ureteral Biomechanics,"Ph.D. thesis, Notre Dame University, Indiana.

[8] Ohlson, L., 1989, "Morphological Dynamics of Ureteral Transport I. Shape and Volume of Constituent Urine Fractions," Am. J. Physiol., 256, pp. R19–R28.

[9] Ohlson, L., 1989, "Morphological Dynamics of Ureteral Transport II. Peristaltic Patterns in Relation to Flow Rate," Am. J. Physiol., 256, pp. R29–R34.

[10] Griffiths, D. J., and Notschaele, C., 1983, "The Mechanics of Urine Transport in the Upper Urinary Tract: The Dynamics of the Isolated Bolus," Neurourol Urodyn, 2, pp. 155–166.

[11] Kim, S., Jacob, J. S., et al (2008) Time-Resolved Dynamic Contrast-Enhanced MR Urography for the Evaluation of Ureteral Peristalsis:Initial Experience Technical report, Wiley-Liss Inc.: New York, pg. 1293 – 1298.

[12] Cox, I. Et al (1992) Ureteric Jets: Evaluation of Normal Flow Dynamics with Color Doppler Sonography Technical report, Medical college of Wisconsin: Milwaukee.

ISBN: 978-988-19252-4-4
ISSN: 2078-0958 (Print); ISSN: 2078-0966 (Online)

[13] Vahidi, B., Fatouraee, N., (2007) a numerical simulation of peristaltic motion in the urete using fluid structure interaction Technical Report, Amirkabir University Of Technology (Tehran Polytechnic): Iran.

[14] Vahidi,B., Fatouraee (2011), ,A mathematical simulation of the ureter : effect of the model parameters on ureteral pressure /Flow relation , Journal of biomechanical Engineening, Vol.133/0310004-1

[15] Vahidi,B., Fatouraee (2012), abimechanical simulation of ureteral flow during presitalsis using intraluminal morphometric data. Journal of theoretical biology, 298(2012)42-50.

[16] Thomas, T. G., Williams, J. J. R. Development of a parallel code to simulate skewed flow over a bluff body [J]. Journal of Wind Engineering and Industrial Aerodynamics. 1997, 67(8): 155-167.

[17]. Munjiza, A. (2004) A Combined Finite/Discrete Element Method. John Wiley & sons Ltd,ISBN 0-470-84199-0.

[18] Munjiza A., Williams JJR, Avital EJ, Cin Ji, Xu D. (2012), Immersed Boundary based Fluid Coupling in Mechanics of Discontinua, Keynote lecture, Advances in Discontinuous Numerical Methods and Applications in Geomechanics and Geoengineering, Edited by Takeshi Sasaki CRC Press/Balkema, DOI *10.1201/b11600-8*

[19] Ed Friedlande ,.(2005) ,Ureter, Urinary bladder , Urethra . [online] Available <http://www.pathguy.com/lectures/bladder> [accessed 11 July 2012]