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Research Article

Rheological response of the human tissues

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A b s t r a c t

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Rupture,
Time constant fluid

The effect of uterus tissue viscoelasticity on its internal pressure is explored. The tissue of the uterus is presented by a linear viscoelastic model with two major time constants. A proper user defined function is developed and incorporated in the simulation software, to represent the model. The geometry of the uterus is separately modeled. It is found that viscoelasticity of the tissue which can be controlled and altered by change the concentration can directly affect its internal pressure. It is also observed that the pressure decreases as the moisture of the tissue is increased. The study is repeated for several practical conditions and parameters pertaining to the viscoelasticity of the tissue are evaluated.

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1. Introduction

Simulation of human tissues, modeling their behavior in different conditions and predicting the parameters influencing their activities are of great interest for medical purposes. A good understanding of mechanical properties of tissues can help protect them against injuries of human organs. There are a great number of studies in literature pertaining to this topic however their experimental backing is scarce. B. Ahn et al [1] studied the deformation and forced response of soft tissues resulting from indentation loadings. In his work the surface deformation of porcine livers is measured both experimentally and numerically. Results show the surface deformation is dependent on indentation depth and tip shapes [1]. Soft tissues have nonlinear, non-homogeneous, anisotropic, and time–velocity dependent behavior [2]. Simulation of large deformations in tissues is a complicated task due to uncertainty of extracted experimental data and complex mechanical behaviors of different tissues in cases involve large deformations [1]. Quantitative responses of soft tissues could be measured experimentally through numerous techniques such as indentation [3-9], aspiration [10-16], shear strain [17-19], and compressive pressure [20-23].

Among different organs the uterus and numerical simulation of its tissue and predicting its behaviors especially for pregnant women is a vital problem. A good understanding of the mechanical responses of the uterus is therefore essential.

In this work the complex behaviors of uterus tissue due to the different internal pressures and its viscoelastic response are simulated numerically.

2. Computational Model

2.1 Geometry of the Model

The uterus of a non-pregnant female is an organ the size of a lemon that is located within the abdominal cavity. During pregnancy the uterus stretches up to the size of a watermelon. [30] The volume increases from 0.005 L to 5 L and as much as 10 L [24]. Thickness of the uterine wall remains uniform prior to delivery and equals to about 1 cm. The uterosacral and round ligaments extend from the uterus to the pelvis and support the uterus. The interior of the uterus contains the fetus surrounded by amniotic fluid and the placenta (Figure 1). The placenta is a vascular organ that acts as a permeable membrane that exchanges oxygen, nutrients, and waste products between the mother and fetus via the umbilical cord. It is a flat, roughly circular structure 2 cm thick in the center. Most placentas, as many as 95%, are in the upper half of the uterus [25].

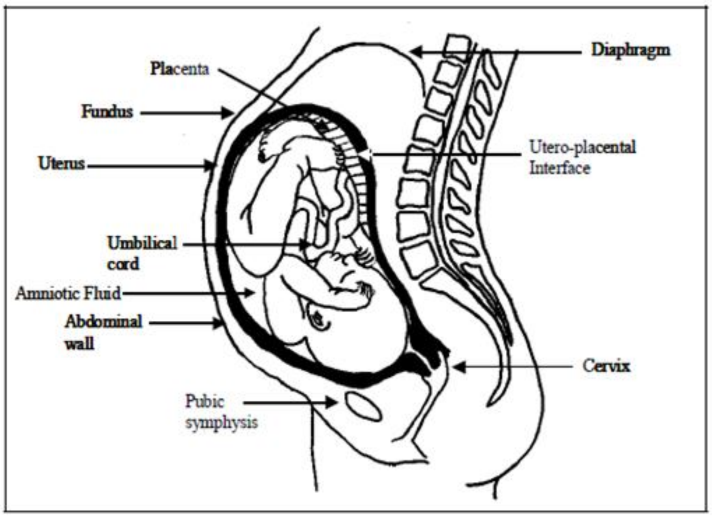

Fig.1: The uterus

A typical uterus has 27 cm long, 18 cm wide and 1 cm thick [30].

In this study a pregnant female with moderate weight in its 36Th week of pregnancy is modelled using Solidworks software as depicted in Figure 2 (c). Uterus front view (a), and side view (b) is used for modeling. Figure 3 shows how two side views of uterus could be used to create 3D geometry of the uterus with the aid of Solidworks.

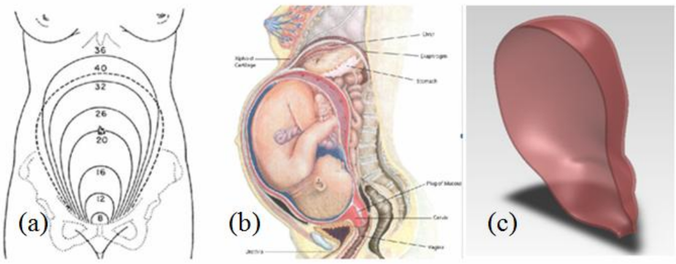

Fig.2 : Uterus front view (a), side view (b), and 3D CAD model (c)

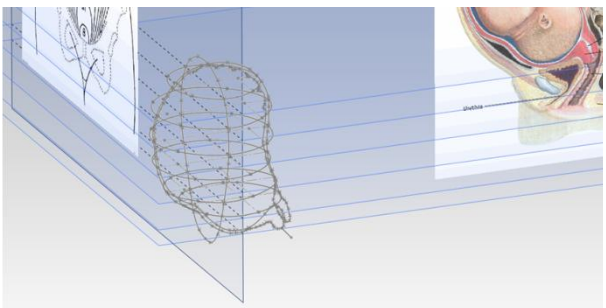

Fig. 3: use of two side views of uterus to create 3D model of the geometry of the uterus with the aid of Solidworks software

Placenta can be modeled as viscoelastic and has anisotropic behaviors [2, 13, & 19]. Tension tests on human uterus tissue have been reported by Pearlman [26], Pearsall [27], and Wood [28]. According to ref. [30] the Young's modulus ranged from 20.3 kPa to 1379 kPa. The average value equal to 566 kPa is implemented for simulation of the elastic linear part of deformation in uterus wall. The Poisson's ratio is set to 0.49 and the density is 1052 kg/m^3 [24].

2.2 Simulation with FEM

Schwartz et al. [7] proposed nonlinear viscoelastic models based on the mass–tensor method and measured the behavior of cervine livers with an in vitro indentation device. Carter et al. [4] measured the Young's modulus of normal and diseased human livers in an in vivo condition. Samur et al. [9] estimated the Young's modulus of the porcine liver as 35–37 kPa and optimized the viscoelastic model parameters with an inverse FEM optimization algorithm.

Table.1: Hyperelastic and viscoelastic Stiffness

	Experiments	Mechanical model	Mechanical Model
Human Uterus	Aspiration	Hyper elastic	Linear Viscoelastic
Stiffness (Pa)		39100	C_{01}=1240 C_{10}=830

The FEM has been widely used in simulations of tissues [24]. In particular, the inverse FEM optimization algorithm [10, 12, 16, 25, 26, and 9] has been applied to tissue property characterization. The algorithm uses FEM simulations iteratively to find the parameters that best fit the experimental results.

In the present study the mechanical behavior is decoupled into a linear viscoelastic stress-relaxation response and a time-independent elastic response [2]. As a linear viscoelastic model, the 2nd order standard linear solid model is expressed as a Prony series expansion in the time domain as follows [1] in section 1.3 of this paper.

2.3 Governing equations

The governing equation is taken from the Prony Model:

$$
\begin{aligned}
G(t) &= k_0 + k_1 e^{-t/\tau_1} + k_2 e^{-t/\tau_2} \\
&= G_0 (1 - \bar{g}_1^p \left(1 - e^{-t/} \right) \quad (1) \quad \left(1 - e^{-t/\tau_2}\right)
\end{aligned}
$$

where k_i, τ_i, t, and $\bar{g}_i^p$ are the rigidity modulus, reduced relaxation time constant, given time, and Prony series parameters, respectively, and can be determined by using the nonlinear least squares method. Tables 1 and 2 lists the implemented viscoelastic and hyperelastic parameters.

Table.2: Rigidity modulus of viscoelastic model

No.	Rigidity modulus(Pa)	Reduced Relaxation time (s)
First	39000	.47
Second	36000	14.7

For the time-independent elastic response, the hyperelastic material model was selected. A hyperelastic material's properties can be determined by the strain energy function (W). Ideally, W is defined with only the parameters required to make a soft tissue model. The incompressible neo-Hookean model, which is widely used in soft tissue modeling, was selected for use here. The strain energy function of the 3D incompressible neo-Hookean model is given by Martijn [27].

$$W = C_{10}(I_1 - 3) \tag{2}$$

Where C_{10} is a mechanical parameter and I_1 is a principal invariant tabulated in table 1.

2.4 meshing and constraints

Figure 4 shows the applied pressure on the internal surface (a), fixed supports at the top and bottom of the uterus (b), and meshed model(c). 3D unstructured grid nominated free tetrahedral volume mesh is used and is refined by adjusting the geometry-based size controller to assure the grid independency of the derived results.

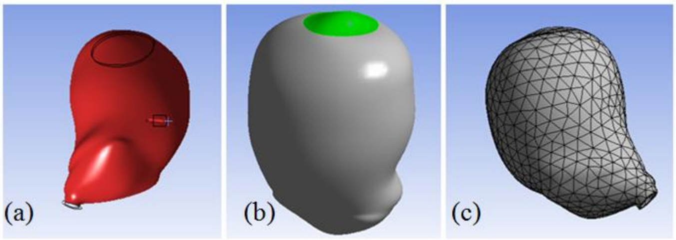

Fig. 4: Pressure on the internal surface (a), fixed supports at the top and bottom of the uterus (b), and meshed model(c)

3. Results and discussion

According to Table 3 the strain value is a vital parameter exceeding some certain values may cause the death of mother or fetus or both of them. The risk of different values is tabulated in Table 3. In this work one percent of minimum value of death risk (44% of death risk) according to ref. [30] is considered as the safe region and is assumed constant. Different runs are implemented for different internal pressures and different values of effective parameters on the visco-elasticity of the tissue. The ratio of rigidity modulus in different cases against common value of rigidity modulus which is tabulated in Table 2 is imposed as the effective viscoelastic parameter. Rigidity module of a viscoelastic is a function of its water contents. As the water contents of the tissue increases its rigidity modulus decreases that is shown in percents of reference values of rigidity modulus.

Figure 5 shows equal deformation of uterus for different pressures inside the uterus and different rigidity modulus ratios. 100% (a), 88% (b), 76% (c), 63% (d), and 114% (e). The rigidity modulus of uterus is according to the datum reported in ref. [1]. As it can be seen the maximum deformation is approximately equal for all cases.

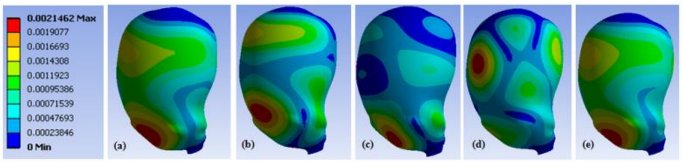

Fig.5: equal deformation of uterus for different pressures inside the uterus and different ratio of rigidity modulus of uterus against reference rigidity modulus. 100% (a), 88% (b), 76% (c), 63% (d), and 114%(e)

Figure 6 shows equal strain of uterus for different pressures inside the uterus and different rigidity modulus ratios. 100% (a), 88% (b), 76% (c), 63% (d), and 114% (e) maximum stress is retained constant and equal for all cases.

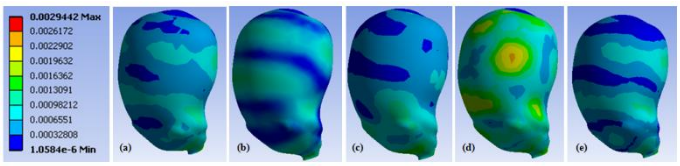

Fig.6: equal strain of uterus for different pressures inside the uterus and different ratio of rigidity modulus of uterus against reference rigidity modulus. 100% (a), 88% (b), 76% (c), 63% (d), and 114% (e)

Figure7 shows different stress distributions of uterus for different pressures inside the uterus and different rigidity modulus ratios 100% (a), 88% (b), 76% (c), 63% (d), and 114% (e). It can be deduced that the maximum stress in viscoelastic tissues with small rigidity modulus is lower. For same strain values this means the tissues with higher moisture contents tolerate lower pressures.

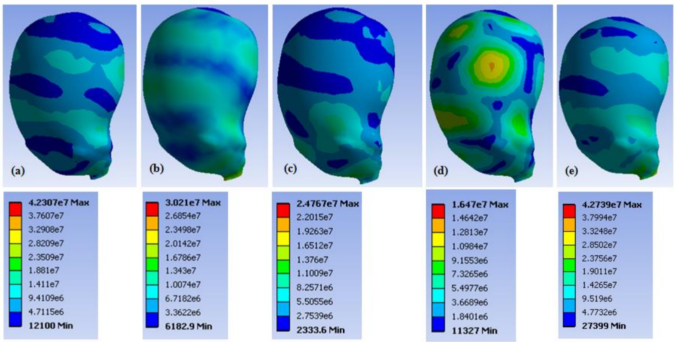

Fig.7: Different stress distributions of uterus for different pressures inside the uterus and different ratio of rigidity modulus of uterus against reference rigidity modulus. 100% (a), 88% (b), 76% (c), 63% (d), and 114% (e)

In Figure 7 the ratio of tolerable pressure to systolic pressure against rigidity modulus ratio for different cases is depicted. According to this graph tolerable internal pressure of the uterus increases as the rigidity modulus ratio increases. In other word as the moisture contents of the tissue increases, the tolerable pressure for the same safe values of the strain decreases.

4. Conclusions

Modeling human tissue is vital in biomedical researches. Analysing the effect of the model parameters is important in the field of surgeries and protections against injuries of human organs. Among different organs the uterus and numerical simulation of its tissue and predicting its behaviors especially for pregnant women is a vital problem. In this study, a pregnant female with moderate weight in its 36'Th week of pregnancy is considered and its uterus is modeled using Solidworks software then the ANSYS software is used for numerical simulation. The mechanical behavior of uterus tissue is decoupled into a linear viscoelastic stress-relaxation response and a time-independent elastic response. The effects of various rigidity modulus ratios on tolerable internal pressures are studied. The maximum safe strain must be retained in a certain level and in all cases this value is considered constant. Simulation results show that the tolerable internal pressure of the uterus increases as the rigidity modulus ratio increases. In other word as the moisture contents of the tissue increases, the tolerable pressure for the same safe values of the strain decreases.

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