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A Viscoelastic Model of a Breast Phantom for Real-Time Palpation

Antoine Widmer and Yaoping Hu, *Member, IEEE*

Abstract— **Palpation of soft tissues helps to diagnose varying diseases within the tissues. Using a phantom, the current method of training palpation lacks for feedback of the training. Similar to a robot-assisted surgical system, a virtual reality (VR) system could be potential for such training due to its interactive nature. In such a VR system, studies revealed the observation that the human perception of objects is insensitive to subtle discrepancies in a simulation. Based upon this observation, we propose a real-time viscoelastic model of a breast phantom (as soft tissues). The model consists of a surface membrane and an inside gel. We evaluate this model through a comparison with a Finite Element Method (FEM) model, featuring physical parameters and different force contacts. The results show that the model can handle multi vertex force contact on an arbitrary location and yields reasonable accurate deformation compared to the FEM model.**

I. Introduction

PALPATION is a diagnostic technique widely used in medical settings. This technique is useful to determine the health of soft tissues, by applying force on the tissues to assess their relative softness through their deformation as visual displacement and force feedback. Currently, training palpation is based upon phantoms (made of various silicones), which lack for feedback to a trainee about the training [1]. Similar to a robot-assisted surgical system [2, 3], a virtual reality (VR) counterpart with a stereoscopic display for rendering visual displacement of soft tissues and a haptic device for providing force feedback of interaction could be useful for training palpation. The fidelity of such a VR system is largely dependent on the model of soft tissues to deform properly under the applied force in real time. Thus, we focus in this paper on developing a real-time viscoelastic model of soft tissues, which mimic the deformation of a breast phantom.

The difficulty of creating a model of a soft tissue is to match the deformation (visual displacement and force feedback) of the tissue with highly viscoelastic properties. In particular, the model needs to govern this deformation in real time, in order to accommodate the human perception of the tissue during user interaction with the tissue. Studies revealed that human perception varies in different settings and is subject to various constraints [4-5]. In our previous studies, we noticed that changing the alignment between a visual display and a haptic device influents human perception when discriminating object softness [4]. We observed that both visual displacement and force feedback of a soft object play crucial roles in this influence. Moreover, O'Sullivan and Dingliana discovered that small inconsistencies in simulating an object do not affect the human perception of the object during real-time interaction [5]. This discovery implies that, when the user interacts with a VR system of training palpation, a real-time model of a soft tissue for the system could carry some margin of deformation discrepancies compared to the actual deformation of the tissue.

Based on these observations, we chose a breast phantom to model in a VR system for training palpation. Being soft and viscoleastic, breast phantoms have the advantage of relatively well studied property parameters. Most research efforts have focused on creating real-time models of soft tissues either with fast rendering, but lack of validation, or with accurate deformation, but lack of rendering speed. These models in general fall into three categories: pure elastic models [6], linear viscoelastic Finite Element Method (FEM) models [7], and nonlinear and pre-computed models [8-9]. Daniulaitis et al. [6] use a pure elastic model of the breast for developing a VR system of learning breast palpation. Pure elastic models suffer from unrealistic deformation, as most soft tissues exhibit both elastic and highly viscous response. To add realism to the simulation, a real-time model used linear viscoelastic governing equation but still does not achieve accurate deformation [7]. Being able to handle large deformation, a non-linear model [8] computes visual displacement of meshed vertices fast enough for real-time visualization. However, it is very slow for rendering real-time force feedback. To speed up computation, a nonlinear model [9] takes an approach of pre-computing in offline the large deformation of soft tissues and using the outcomes of this precomputation to render visual displacement and force feedback in real time. Although the approach simulates well predefined deforming scenarios of soft tissues, the pre-computation is not able to cope with unexpected scenarios.

Although physical breast palpation systems exist [10], we are not able to find reports on modeling a viscoelastic breast phantom for rendering real-time deformation. Derived from the idea of a surface membrane with an internal equation of state as in [11], we modify the surface membrane and equation of state with viscoelastic features. These modifications allow the current model to simulate important characteristics of the breast phantom such as its volume conservation and viscoelastic response. Because the current architecture of personal computers could not permit a model of soft tissues to achieve both real-time computation and 100% accurate deformation [12], our model takes an approach of optimizing its computational speed at some cost

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A. Widmer and Y. Hu are with the Dept. of Electrical and Computer Engineering, University of Calgary, Calgary, AB, CANADA (e-mail: widmera@ucalgary.ca and huy@ucalgary.ca).

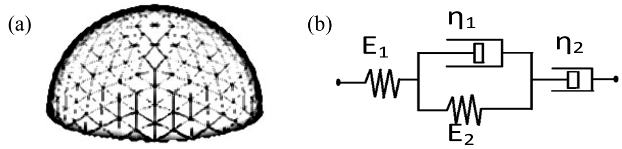

Figure 1. Surface of the real-time model: (a) general aspect, and (b) Burger model governing viscolestic link between surface vertices.

of reduced accuracy of deformation when compared to its FEM counterpart (as a reference). Against this reference, the accuracy of our model of the same soft tissues could be assessed. This approach is feasible due to the observation that the human perception is insensitive to subtle discrepancies during real-time simulation [6]. Thus, our model is able to compute deformation of the breast phantom in terms of its visual displacement and force feedback in real time – every 10 ms.

Using our evaluation method [11], we validate our current real-time model by comparing the deformation of meshed vertices (in terms of both visual displacement and force feedback) to that of the same vertices governed by a FEM counterpart. Both real-time and FEM models have same geometry and are considered to contain the same soft tissues. The results reveal that, under the real-time model, visual displacement of all vertices and force feedback of the vertices within the area of finger contact are in good agreement with those under the FEM model. This indicates that our real-time model is potential to describe the deformation of the breast phantom during palpation.

We organize this paper as follows: Section II describes the two components of the real-time model, the value of each real-time parameter to simulate a breast phantom and the method we used to evaluate the model; Section III presents the evaluation results followed by a brief discussion. Section IV gives our conclusion and future work.

II. METHODOLOGY

Fig. 1(a) illustrates the meshed shape of a real-time model mimicking a breast phantom. This mesh is the same as the one used in our previous work (8.0 cm in diameter) [11]. However the governing equations to simulate the breast phantom deformation are updated to take account the high viscoelasticity of the breast phantom. The real-time model consists of a mesh of vertices as a surface membrane and a state equation simulating an uncompressible and viscoelastic material as an inside gel. With no internal vertices, the model includes 338 surface vertices as interactive vertices where contact forces can be applied.

A. Surface Membrane

Departing from our previous work [11], each vertex of the surface membrane is linked to its neighbors through an assembly of dashpots and springs. Known as a Burger's material, this assembly includes a Maxwell material and a Kelvin material in series as illustrated in Fig. 1(b). The Burger's material has the advantage of exhibiting loading and unloading time-independent strain and time-dependent recovery similar to a soft tissue such as a breast [13]. Therefore, the Burgers material allows the real-time model to simulate a real breast phantom surface membrane. We used the following equation to describe the behavior of the surface membrane [13]:

$$\sigma + \left(\frac{\eta_1}{E_1} + \frac{\eta_1}{E_2} + \frac{\eta_2}{E_2}\right)\dot{\sigma} + \frac{\eta_1\eta_2}{E_1E_2}\ddot{\sigma} = \eta_1\dot{\varepsilon} + \frac{\eta_1\eta_2}{E_2}\ddot{\varepsilon}, \quad (1)$$

where σ is the stress of the model, $\dot{\sigma}$ and $\ddot{\sigma}$ represent the first and second time derivative of the stress respectively. η_1 and η_2 represent the two dashpots and E_1 and E_2 represent the stiffness of the two springs. $\dot{\varepsilon}$ and $\ddot{\varepsilon}$ represent the first and second time derivative of the strain respectively. To implement the Burger material, we discretized Eq.(1) by assigning a weight (vertex's weight) to each vertex on the surface and an initial distance (rest length) between two vertices at rest. To compute the strain and stress at each current time step, we used data from the previous time step.

B. Inside Gel

Our real-time model does not contain any link or vertex inside the membrane. Therefore, a state equation is required to keep the shape of the breast phantom consistent. Removing internal vertices is to accelerate the computation of the real-time model. As a starting point we use the ideal gas equation described in our previous work [11]. However, this ideal gas equation does not simulate viscoelasticity. Therefore, we modified the ideal gas equation and the algorithm to reflect the viscoelasticity requirement. At the initial step, we compute the internal force needed to keep the volume stable with the following equation:

$$F_{p,0} = \frac{P}{V}, \quad (2)$$

where $F_{p,0}$ is the force keeping the shape of the phantom stable at time 0. P represents the pressure inside the phantom and V is the initial volume of the phantom computed through the divergence theorem. For the following frames, the internal force is computed through the following state equation:

$$F_{p,i} = F_{p,i-1} - F_{p,i-1} * \frac{V_i - V_{i-1}}{V_{i-1}} * a_1 - a_2 * V/dt + a_3 * \dot{F}/dt \quad (3)$$

where $F_{p,i}$ is the force keeping the shape of the phantom stable at frame i >1. $F_{p,i-1}$ and V_{i-1} represent the internal force and the volume of the phantom at frame i-1, respectively. a_1, a_2 and a_3 are empirical factors to tune the force for keeping the volume of the phantom constant. dt is time step and $\dot{F}$ represents the first derivative of the force keeping the shape of the phantom stable. With the combination of surface membrane and inside gel, the real-time model can simulate a breast phantom.

C. Parameters

To simulate a breast phantom as soft tissues, we need to approximate the real-time parameters from the physical world. Two different studies provide viscoelasticity and hyperelasticity physical parameters, respectively [14, 15].

Viscoelasticity parameters are derived from a report extracting in-vivo breast viscoelasticity from ultrasonic measurement [14]. The study finds that the breast shows linear viscoelastic behavior up to 5N of exerted force. From the collected data, we are able to extract a two term Prony series [13] as follows:

$$C_{ij}^{R}(t) = C_{ij}^{0}(1-\sum_{k=1}^{N} g_k^P (1-e^{-\frac{t}{\tau_k}})), \tag{4}$$

where g_k^p and τ_k are the k^{th} Prony constants and the k^{th} Prony retardation time constants, respectively; t is the current time-step and C_{ij}^{0} is the Neo-Hookian hyperelastic parameter. N is the number of terms (N = 2 for our case). However, the Phony series yields a recovery time (time needed to have the phantom back in its original shape) significantly longer than its counterpart found from an actual breast phantom, when using a finger to press/release this actual phantom. We reduce the recovery time to 1.0 sec according to our observation while pressing/releasing the physical phantom by the finger. Hyperelastic parameters are obtained from a report describing Neo-Hookian parameters of a breast phantom [15]. The Neo-Hookian equation in Eq. (5) governs the surface membrane and inside gel hyperelastic components [15]:

$$U = C_{10}(\bar{I}_1 - 3) + \frac{1}{D_1}(J^{el}-1)^2, \tag{5}$$

where U represents the strain energy per unit of reference volume; C_{10} and D_1 are material parameter; J_{el} is the elastic volume ratio I_1. Table I and II indicate the proper values for the modified Prony serie and Neo-Hookian parameters for membrane and inside gel respectively.

Because the real-time model uses fast computation algorithms, we need to approximate its parameters from the deformation yielded by the parameters from the literature. To achieve this, we create a Finite Element Method (FEM) model. As illustrated in Fig. 1, the FEM model shares the same outside geometry as the real-time model. The FEM model includes surface membrane and insidel gel. Both materials use a two-term Prony series to implement a viscous component. Governed by Eqs. (4) and (5), the FEM model uses the parameters from Table I and II. As comparison, we define a step-wise force profile lasting 4 seconds (s.). This force profile includes a force of 3 N (maximum force sustained for a longer period of time by the haptic device) during 2 s. without ramping followed by 2 s. without force. From this simulation, we record the top vertex displacement. Based on this displacement, we manually update the different parameters until the displacement of the top vertex governed by the real-time model mimics its counterpart governed by the FEM model. Table III and IV show the parameters to simulate the phantom in real time.

On a workstation (DELL Precision 690 with 2 dual-core processors at 3.2GHz and 4 GB of RAM), the model is able to compute the visual displacement and force feedback of every vertex in about 10 ms for each time step allowing real-time interaction for human user.

D. Evaluation method

This evaluation method includes two steps (data acquisition and data processing) [11]. The data acquisition step consisted of different activities such as creating a golden standard for comparison, defining force profile, applying distributed forces and recording data. As a golden standard, we use the same FEM model as described in the previous subsection (C. Parameters). FEM has the advantage of solving mechanical models with great accuracy, but has difficulties in solving models in real-time. On the same computer used for the real-time model simulation, the FEM model needs 22.5 s for each time step

To verify the deformation of the real-time model in a different way than introduced in [11], we select a randomly off centered contact area for applying the same force profile as introduced in subsection Parameters. This force profile is used in four different force distributions. Distribution I features a single point contact; Distribution II features a multi-vertex contact area and Distribution III features a 2D Gaussian distribution with the peak on the central vertex of the multi vertex contact area. Distribution IV features a 2D Gaussian distribution with the peak on an off-centered vertex of the multi vertex contact area. This distribution represents the inclination of the finger to use its distal section for palpating as normally observed. Distribution II to IV shared the same contact area of 5.2 cm^2. We recorded displacement on every vertex on the membrane and force feedback only on vertices in contact. As data processing step, we used a statistical method taking account the human

TABLE I
SURFACE PARAMETERS EXTRACTED FROM LITTERATURE

Thickness	2 mm			
Density	950 kg/m^3			
Neo-Hookian parameters	C_{10}	700.3 kPa	D_1	0.001
Prony series	g_1^p	*0.9*	τ_1	0.002 s
	g_2^p	0.09	τ_2	0.005 s

TABLE II
INSIDE GEL PARAMETERS EXTRACTED FROM LITTERATURE

Density	950 kg/m^3			
Neo-Hookian parameters	C_{10}	10.3 kPa	D_1	0.001
Prony series	g_1^p	0.9	τ_1	0.002 s
	g_2^p	0.09	τ_2	0.005 s

TABLE III
SURFACE PARAMETERS FOR THE REAL-TIME MODEL

Burger's model parameters	E_1	0.6 N/cm
	E_2	0.01 N/cm
	η_1	0.49 N*s/cm
	η_2	0.01 N*s/cm
Implementation parameters	Vertex weight	0.001 kg
	Rest length	30%

TABLE IV
INSIDE GEL PARAMETERS FOR THE REAL-TIME MODEL

Pressure	P	100 N/mm^2
Modifying factors	a_1	120
	a_2	0.01
	a_3	0.001

TABLE V

COMPARISON OF VERTICES DISPLACEMENT GOVERNED BY THE REAL-TIME MODEL WITH THAT UNDER ITS FEM COUNTERPART.

	RMSE [cm]	One-way ANOVA		Bland and Altman agreement	
		F	***p***	**SD [cm]**	**Agree [%]**
Distr. I	0.17	2.37	0.12	0.27	95.2
Distr. II	0.21	1.52	0.36	0.29	95.0
Distr. III	0.26	1.69	0.32	0.32	95.3
Distr. IV	0.32	0.96	0.48	0.33	96.5

TABLE VI

COMPARISON OF FORCE FEEDBACK GOVERNED BY THE REAL-TIME MODEL WITH THAT UNDER ITS FEM COUNTERPART.

	RMSE [N]	One-way ANOVA		Bland and Altman agreement	
		F	***p***	**SD [N]**	**Agree [%]**
Distr. I	0.05	0.12	0.86	0.01	98.6
Distr. II	0.12	1.2	0.31	0.12	96.1
Distr. III	0.15	1.93	0.29	0.21	95.2
Distr. IV	0.16	2.02	0.21	0.25	95.1

perception to assess a real-time complex model of a soft tissue [11].We computed RMSE, used an ANOVA analysis and performed a Bland and Altman Agreement method for both vertex displacement and force feedback [16]. Both ANOVA analysis and the agreement method must have a *p*-value of more than 0.05 and an agreement of more than 95% of the data within ±2 standard deviation (SD) in each distribution to be successful.

III. RESULTS AND DISCUSSION

Table V and Table VI show the comparison results for vertices displacement and force feedback respectively. For all distributions, more than 95% of the data are in agreement for both vertices displacement and force feedback; and their *p*-values from the ANOVA analysis are well above the 0.05 threshold set in our evaluation method. Based on the observation that humans are not sensitive to small inconsistencies in a VR simulation [5], the real-time model could be possibly sufficient for user interaction in training palpation. Nevertheless, further studies on humans are needed to verify this sufficiency.

As shown in Table V, all distributions have similar level of agreement with SD values increasing from distribution I to IV. Comparatively to this observation, SD increases as well from distribution I to distribution IV in force feedback comparison, as illustrated in Table VI. However a general decreasing trend is apparent for agreement ratio when comparing force feedback, with a noticeable difference between the single-vertex contact (Distribution I) and multi-vertex contact (the other distributions). This noticeable difference reflects the varying numbers of data points for the comparison of force feedback under Distribution I (1 data point in 400 time steps) and the other distributions (20 data points in 400 time steps). However the agreement data decrease. This observation is due to the fact that there is less data above the threshold even if a larger portion (±2 SD) is considered. This shows the increase of contacts' complexity brought by Distribution III and IV for force computation.

In summary, every distribution reaches a level of agreement of more than 95%. Therefore the implementation of the real-time is possible for user interaction. This successful example paves the road for simulating other internal soft tissues. However, studies are needed to verify if human users perceive the deformation of the real-time model correctly.

IV. CONCLUSION

In this current work, we proposed a new method to develop real-time models of soft-tissues for multi-vertex contact of palpation. Future work includes human studies to verify whether the real-time model is sufficient to the user interaction within a VR system.

REFERENCES

[1] W.M. Lam, C.P. Chan, C.F. Chan, and M.H. Tang, "Factors affecting the palpability of breast lesion by self-examination," *SMJ*, vol. 49, pp. 228-232, 2008.

[2] L.J.M. van den Bedem, R. Hendrix, G.J.L. Naus,R. van der Aalst, P.C.J.N. Rosielle, H. Nijmeijer, J.G. Maessen, I.A.M.J Broeders, and M. Steinbuch "Sofie, a robotic system for minimally invasive surgery", *Proceedings of the 6th International MIRA Congress*, May 2011, Athens, Greece, pp. P056, 2011

[3] D. F. Louw, T. Fielding, P. B. McBeth, D. Gregoris, P. Newhook, G. R. Sutherland. "Surgical Robotics: A Review and Neurosurgical Prototype Development." *Neurosurg*, vol. 54, pp. 525-537, 2004.

[4] A. Widmer.and Y. Hu; , "Effects of the Alignment Between a Haptic Device and Visual Display on the Perception of Object Softness," *IEEE Trans. on SMCs, Part A*, vol.40, no.6, pp.1146-1155, Nov. 2010

[5] C. O'Sullivan and J. Dingliana, "Collisions and perception," *ACM Trans. Graph*., vol. 20, no. 3, pp. 151–168, 2001.

[6] V. Daniulaitis, M. Alhalabi, H. Kawasaki and Y. Tanaka, "Medical palpation of deformable tissue using physics-based model for haptic interface robot (HIRO)," *Proc. IROS'04*, vol.4, pp. 3907-3911, 2004.

[7] I. Brouwer and V. Mora and D. Laroche, "A viscoelastic soft tissue model for haptic surgical simulation", WHC'07, pp. 593-594, 2007.

[8] J-M. Schwartz, M. Denninger, D. Rancourt et al., "Modelling liver tissue propetries using non-linear visco-elastic model for surgery simulation." *Medical Image Analysis*, vol. 9, pp. 103-112, 2002

[9] E. Basafa and F. Farahmand, "Real-time, simulation of the non-linear visco-elastic deformations of soft tissues", *Int J CARS*, 2010 .

[10] L. Chaitown, *Palpation and Assessment Skills,* Elsevier , 2003

[11] A. Widmer and Y. Hu, "Statistical comparison between a real-time model and a FEM counterpart for visualization of breast phantom deformation during palpation," *Proc. CCECE 2010*, Calgary, AB, Canada, 2-5 May. pp 1-4, 2010.

[12] R. Stuart, *Design of Virtual Environments*, Barricade Books, 2001.

[13] W.N. Findley, J.S.Y. Lai, and K. Onaran, *Creep and relaxation of nonlinear viscoelastic materials, with an introduction to linear viscoelasticity*, American Elsevier Pub. Co., 1976.

[14] M. Sridhar and M.F. Insana, "Ultrasonic measurements of breast viscoelasticity", *Med. Phys*., vol 34, pp. 4757-4767, 2007.

[15] H. Mehrabian and A. Samani, "An iterative hyperelastic parameters reconstruction for breast cancer assessment", *Proc. of the SPIE, Medical Imaging 2008*, pp. 69161C-1-69161C-9, 2008.

[16] J.M. Bland and D.G. Altman, "Statistical methods for assessing agreement between two methods of clinical measurement," *Lancet*, vol. 1, pp. 307–310, 1986.