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A mass-spring-damper model for real time simulation of the frictional grasping interactions between surgical tools and large organs

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Abstract. Considering the loss of direct visual and tactile information, surgeons require special training programs to obtain sufficient proficiency for laparoscopic surgery. Surgical training simulation systems provide an effective alternative to animal models for repetitive training practices. The purpose of this study was to develop a biomechanical model of large soft organs for simulation of the interactions of a surgical grasper and spleen in real-time. The mechanical behavior of the spleen was molded in detail, including its nonlinear hyper viscoelastic properties using a mass-spring-damper model. A novel collision detection algorithm was used to determine the tool-tissue contact zones. Force-based and geometry-based boundary conditions were imposed at the contact nodes, respectively, to represent slippage-included and slippage-free grasping conditions. The model's predictions were validated against the experimental results on a synthetic test sample. Results of simulation of interactions between the grasping tool and the spleen organ indicated that the non-linear rate dependent and stress relaxation behaviors of the tissue was well depicted by the model. Also, the model was capable of reflecting the effect of tool-tissue friction coefficient on the slippage-free or slippage-included grasping behaviors.

1. Introduction

Minimally Invasive Surgery (MIS) is increasingly recognized as an effective alternative to traditional open surgery. MIS operations on the internal abdomen organs are performed as laparoscopic surgery in which a miniature video camera and long narrow surgical instruments are inserted into the abdomen cavity through small incisions. The camera provides an image of the interior of the abdomen, enabling the surgeon to explore the internal organs and perform the operation using the surgical instruments.

Laparoscopic surgery has significant advantages over open surgery. It causes less operative trauma and post-surgical complications that would shorten the hospitalization time and cost. Also, it permits much faster recovery for the patient, which is of great physiological and psychological importance [1]. However, it is a more difficult job for the surgeon and usually takes longer than conventional open surgery [2]. The 2-D image provided by the camera does not contain the depth information needed to manipulate organs effectively. Furthermore, the surgeon lacks the direct line-of-sight visualization and has difficulty in handling the

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instruments due to their limited maneuverability and the fulcrum effect, i.e. movement of the instrument's tip in the opposite direction of the surgeon's hand [3].

Considering the loss of direct visual and tactile information, surgeons require to go through special training programs to obtain sufficient proficiency for laparoscopic surgery. This program would include repetitive practices for improving hand-eye coordination, as well as getting adequate skill for working with instruments with limited maneuverability under fulcrum effect [4]. In the past, the laparoscopic surgery training was mostly conducted in animal labs on animal models. This method, however, is restricted by the fact that it needs special housing and facilities and could not be practiced repetitively due to ethical and economic problems. An alternative approach is to use surgical simulation systems, which provide a virtual environment to replicate the real surgical conditions [3]. Such systems, in general, include two main components:

1. A virtual reality based interactive graphical environment that provides a realistic simulation of the mechanical interactions between the tool and organs;
2. A force feedback device that transfers the force interactions to the surgeon's hand [5].

An essential requirement of surgical simulation systems is functioning in real time. The system should be able to simulate the tool-tissue interactions with an update rate of at least 30 Hz for graphical rendering [6] and about 1 kHz for driving the haptic device [7]. This is obviously a challenging task considering the complicated mechanical behavior of the human soft tissues, including rate dependency and nonlinearity, either due to their inherent material properties [8,9] or large deformation behavior [5]. Several approaches have been proposed for modeling of deformable objects in real-time surgery simulation. The non-physical models, e.g. free-form deformation [10] and deformable spline [11] are based on pure mathematical representation of the organ's geometry and do not provide a simulation of its mechanical behavior. The physics-based approaches include a vast range of methods from continuum based methods, e.g. finite element [8,9], boundary element and linked volume modeling, to discrete methods, e.g., Mass-Spring-Damper (MSD) [5,12,13] and meshless modeling [6,14]. In contrast to non-physical models, physics-based approaches take the mechanical characteristics of the organ into account.

There have been few studies in the literature attempting to model the tool-tissue interactions of large intra-abdominal organs, e.g. spleen. In the work of Tirehdast et al. [15,16], the simple interactions of spleen tissue with a surgical instrument was modeled using the finite element method. However, their model did not run in real-time due to the high computational cost. In the work of Abdi et al. [14], the meshless element-free Galerkin method was generalized to develop an algorithm capable of dynamic simulation of the grasping procedure of large organs. Although the mechanical model was rather complicated, taking the viscoelasticity of the organ into account, its interactions with the grasper were simplified as a simple compressive loading. Also, in order to obtain a real-time performance, the number of nodes of the model was reduced that affected the accuracy of the results.

The purpose of the study was to provide a model for simulation of the interactions of spleen tissue and a surgical grasper in real-time. The mechanical behavior of the spleen is molded in detail, including its nonlinear viscoelastic properties, using an MSD model. Also, the grasper is considered to have a complex structure, with three paralleled fingers. Finally, different grasping modes, including the possibility of a sliding response, are simulated by incorporating different friction coefficients between the tool and tissue.

2. Method

2.1. Modeling

A biomechanical model was developed to simulate the interactions of the spleen tissue with a specially designed three-jaw large organ grasper [17,18] (Figure 1). The spleen tissue was modeled as a deformable object with nonlinear viscoelastic properties. The 3D geometry of the spleen was obtained from the CT-scan images of a human subject. The CT slices were processed using ITK-SNAP segmentation software (www.itksnap.org) to obtain the tissue boundaries. The 3D geometry of the spleen was then reconstructed in SOLIDWORKS (Dassault Systèmes, Massachusetts, USA), and meshed into polyhedrons by NETGEN mesh generator (SourceForge, Dice Holdings, San Francisco, USA). The vertexes and edges of these polyhedrons were considered as the mass points and the interconnected springs, respectively, of a Mass-Spring-Damper (MSD) model of the tissue. The jaws of the grasper were modeled as rigid links that always move in parallel, considering the parallelogram mechanism implemented in the

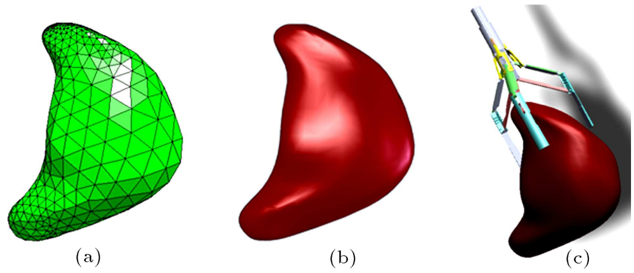
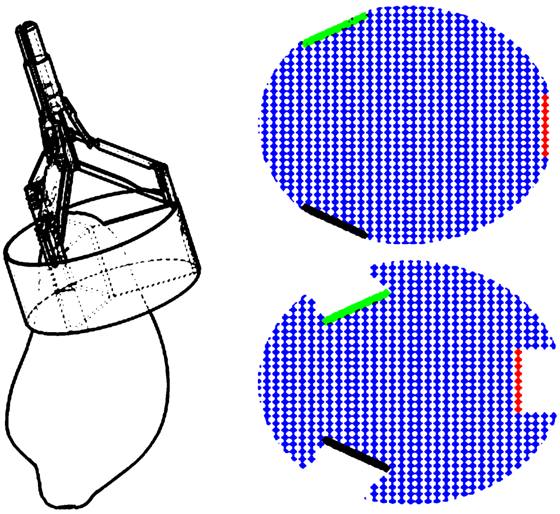

(a) (b) (c)

Figure 1. The meshed (a) and rendered (b) spleen tissue in interaction with a three fingered large organ grasper (c).



(a) (b)

Figure 2. The collision detection algorithm. A hypothetical hollow cylinder was used to describe the configuration of the grasper jaws (a). The nodes on the tissue with coordinates consistent with those of the jaws were considered as contact nodes and attached to the jaws kinematically.

grasper [17]. Each jaw had a planar inferior surface, with fine ribbed profiles that were neglected in the model.

A special collision detection algorithm was designed and implemented to speedup finding the points at which the contact between jaws and tissue occurs (Figure 2). The relative configuration of the three jaws of the grasper could be described using a hollow cylinder in which the diameter decreases when the jaws are being closed. Based on this simple model, the contact between jaws and tissue occurs only if a node on the surface of the tissue crosses through the cylinder surface. In order to implement this algorithm, the Cartesian coordinates of the nodes on the surface of the tissue were converted into cylindrical coordinates and transferred so that the axis of the cylindrical coordinate system coincides with the grasper axis. Then, the radius distances of the nodes from this axis were calculated, enabling the nodes that could potentially contact with any of the three jaws to be identified in a single step. In the next stage, only the nodes, whose angular coordinates, i.e. azimuth, were consistent with those of the jaws, were selected as the contact nodes. Interaction of these contact nodes with the jaws, in fact, determined the boundary condition of the MSD model of the tissue.

In a general MSD model, the positions of nodes are calculated through a differential equation system, so that the force balance is maintained for all nodes in presence of either a force or a geometrical boundary condition. For our spleen-grasper model, the contact nodes at the tissue are limited to move only in the plane of their corresponding jaws (Figure 2). If the external pull force is larger than the tangential component of the force between the jaw and tissue, slippage would happen. In this case, a force constraint might be applied to the nodes to represent the friction force. On the other hand, if the tangential forces are smaller than the external pull, the contact nodes stick to the jaw with no slippage in between. In this case, the grasper jaws would impose their kinematics, e.g. displacement, velocity and acceleration, to the corresponding contact nodes. Thus, in general, two kinds of force and geometrical boundary conditions could be used in our model to represent a grasping procedure with and without slippage, respectively.

The mass-spring-damper model of the large organ consisted of discrete mass nodes distributed throughout the organ and interconnected via a network of springs and dampers. For each mass node [19] we have:

$$m_i\ddot{r}_i = F_i^{\text{spring}} + F_i^{\text{damping}} + F_i^{\text{external}}, \tag{1}$$

where F_i^{spring} represents the total elastic force applied to node i from the adjacent nodes, and F_i^{damper} and F_i^{external} are the global damping and external forces applied to node i, respectively. Also, $\ddot{r}_i$ is the acceleration vector for mass m_i. This equation might be extended to:

$$m_i\ddot{r}_i = \sum_{j=1}^{n} F_{ij}^{\text{spring}}(\Delta l_{ij})\frac{r_j - r_i}{\|r_j - r_i\|} + F_i^{\text{damping}}(r_i, \dot{r}_i) + F_i^{\text{external}}, \tag{2}$$

where:

$$\Delta l_{ij} = (r_j - r_i) - (r_{oj} - r_{oi}). \tag{3}$$

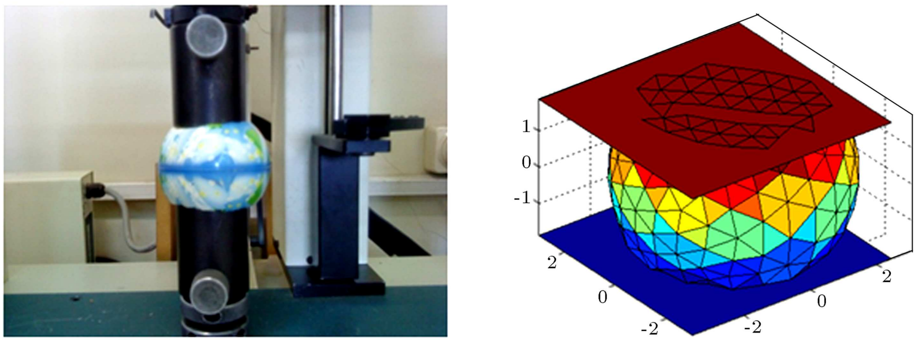
In the above equations, r_i and r_j are the current and r_{oi} and r_{oj} are the initial position vectors for nodes i and a typical adjacent node j, respectively. Also, $\dot{r}_i$ represents the current velocity vector for node i, and n is the total number of its adjacent nodes. Considering the nonlinear elastic behavior of the soft living organs, we used a two-step expression for the force-displacement characteristics of the spring [4]:

$$F_{ij}^{\text{spring}}(\Delta l_{ij}) = \left\{\begin{array}{ll} K_1\Delta l_{ij} + K_2\Delta l_{ij}^3 & \Delta l_{ij} \leq \Delta l_c \\ (A+B(|\Delta l_{ij}|-|\Delta l_c|))\text{sgn}(\Delta l_{ij}) & \Delta l_{ij} \geq \Delta l_c \end{array}\right\}, \tag{4}$$

where K_1 and K_2 are constants, Δl_c represents the critical displacement below which the springs show nonlinear behavior, and parameters A and B are defined as follows:

$$A = K_1\Delta l_c + K_2\Delta l_c^3, \tag{5}$$

$$B = K_1 + 3K_2\Delta l_c^3. \tag{6}$$



(a) (b)

Figure 3. The mechanical properties of the test sample were obtained using unconstrained compression tests (a), and simulation of the same boundary and force conditions in the MSD model (b).

Also, in order to simulate the non-linear viscoelastic behavior of the soft biological tissues, the damping force was assumed to consist of a displacement-velocity component, in addition to the typical velocity alone component, to imply the combined effect of strain and strain rate [4]:

$$F_i^{\text{damping}}(r_i, \dot{r}_i) = b_0 \dot{r}_i + b_1 \|r_i - r_i^0\| \dot{r}_i, \tag{7}$$

where b_0 and b_1 are two damping constants and r_i^0 represents the rest position of node i. It should be noted that, in general, the parameters of the model, e.g. m, K_1, K_2, b_0 and b_1, can be different for each individual spring and node. However, due to the fact that determination of a very large number of model parameters is not practical, this formulation was reduced to include constant parameters for the entire model, assuming a homogenous and isotropic mechanical behavior for the spleen tissue.

In the next stage, a parameter tuning approach was implemented to obtain the relationship between the parameters of the model and the mechanical properties of the tissue under consideration. In order to determine the model parameters, an optimization procedure was followed, with the objective function defined as the sum of the squared difference between the experimental data points and the corresponding simulated responses of the model, when subjected to the same loading and boundary conditions. The Fmincon solver of MATLAB Optimization Toolbox (MATLAB, Mathworks Inc., Natick, Massachusetts) was used to minimize the objective function and obtain the model parameters best fitting the experimental behavior.

In order to update the positions of the nodes, it was required to integrate the 2nd-order ordinary differential equations system (2), simultaneously. Among the different implicit and explicit numerical integrating approaches available, we used the Finite Difference Method due to its high computational speed and acceptable accuracy.

3. Experiments

The proposed model was validated using experiments on a simple spherical polymeric test sample. At first, the mechanical properties of the test sample were obtained from unconstrained compression. The test sample was located between the jaws of a Hounsfield test machine and was loaded at different rates of 6 mm/min and 60 mm/min (Figure 3(a)). The optimization procedure, as described above, was used for tuning the parameter of the MSD model of the test sample, while simulating its behavior in a similar condition to that of the experimental study (Figure 3(b)).

In the next stage of the experimental study, the mechanical behavior of the test sample, during grasping, was investigated (Figure 4). An instrumented three fingered large organ surgical grasper [17,20] was used to manipulate and grasp the test sample under known compressive forces at the jaws. During the experiment, the test sample was grasped by the device with an increasing pinch force, recorded through the tiny self-temperature compensated strain gages (BFLA-2-8, TML Co., Ltd., Tokyo, Japan) attached to the jaws. Meantime, the position data of the encoder of the system's servo DC motor (MR, Typ. M, Maxon Motor AG., Switzerland) was recorded to determine the displacement of the jaws, using the kinematics equations of the instrument [17]. The results were

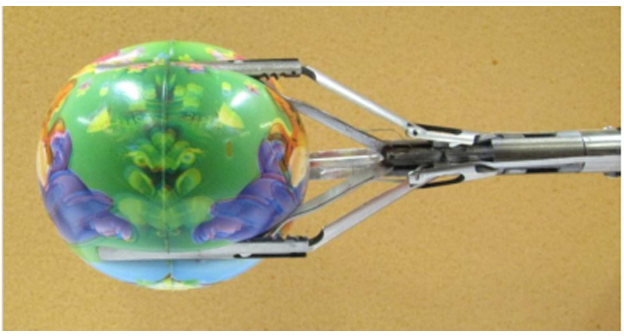


Figure 4. The spherical polymeric test sample while grasping with the three fingered laparoscopic grasper.

used to determine the pinch force-deformation behavior of the test sample during three fingered grasping and were compared with the predictions of the model while simulating the same procedure.

4. Result

The results of the unconstructed compression tests of the synthetic test sample at two different loading rates are illustrated in Figure 5. The higher stiffness of the test sample at 60 mm/min loading rate, in comparison with that of the 6 mm/min, is an indicative of its viscoelastic properties. The MSD model parameters of the test sample, obtained using the parameter tuning procedure are indicated in Table 1.

Table 1. The MSD model parameters of the test sample.

Model parameter	Value
m (kg)	1
K_1 (N/m)	0.5
K_1 (N/m^3)	1
b_0 (N.s/m)	2
b_1 (N.s/m^2)	0

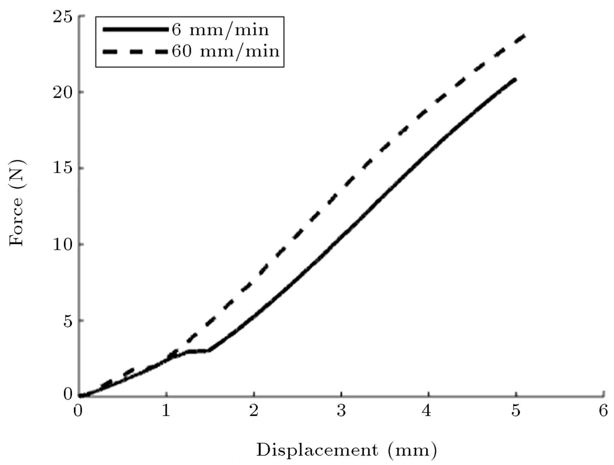


Figure 5. The force-deformation behavior of the test sample during unconstrained compression tests.

The results of the experimental study performed to validate the modeling approach of the present study are compared with those of the simulation in Figures 6 and 7. In general, the model predictions for the deformations of the test sample were in good agreement with the results of the experimental study, indicating similar geometries and curvatures under the jaws and in non-contact areas (Figure 6). Also, there was a very good correlation (correlation coefficient of 0.98) between the force-deformation behavior of the test sample predicted by the model and the experimental data (Figure 7). This was particularly true for the low and middle ranges of sample deformation; the deviation between the forces predicted by the model and measured experimentally increased at large deformations up to a maximum of 2.6 N.

The predictions of the model for deformation of the spleen tissue during grasping with the instrument are shown in Figures 8 and 9. It was assumed that the grasper closed for 5 seconds, then stopped moving for 7 seconds, and then started to pull the tissue. The coefficient of friction between the jaws and tissue was assumed to be 0.4, 0.6 and 0.8. Results indicated that the model was able to mimic the mechanical

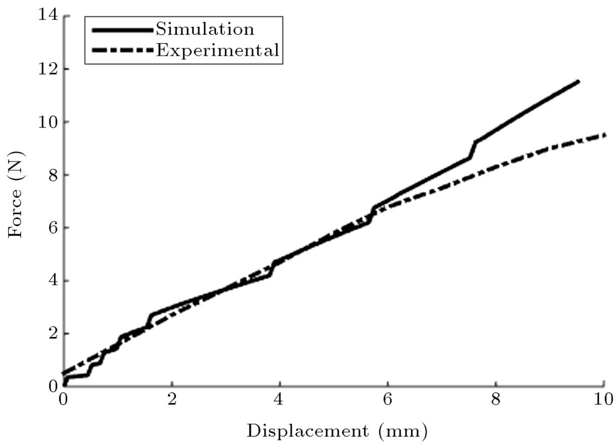


Figure 7. The force-deformation behavior of the test sample predicted by the model in comparison with the results of experimental test.

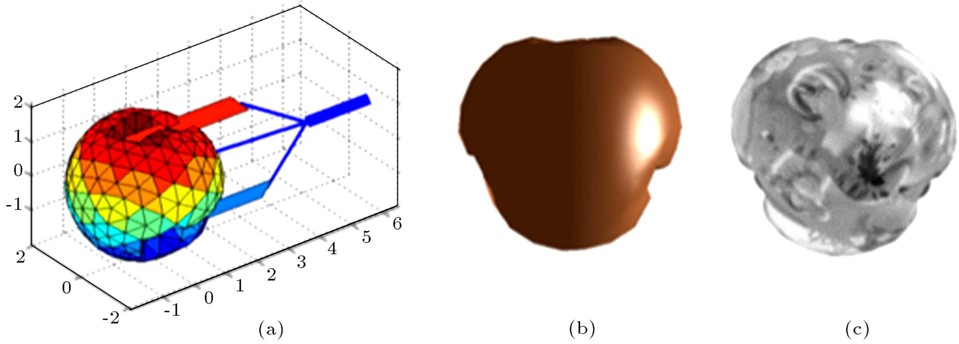


Figure 6. Deformations of the test sample predicted by the model ((a) and (b)) and observed experimentally at an identical grasping force.

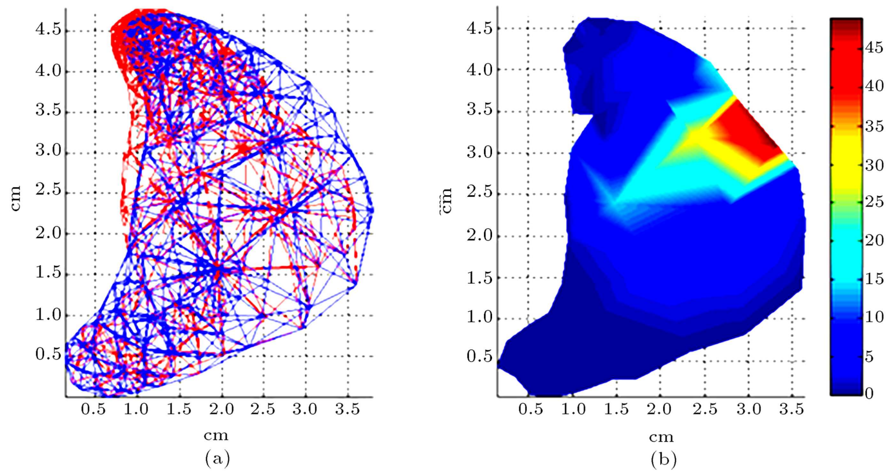


Figure 8. (a) The deformation of spleen tissue predicted by the model before and after grasping. (b) The normal stress distribution in the lateral cross section of spleen tissue at initial grasping is also illustrated.

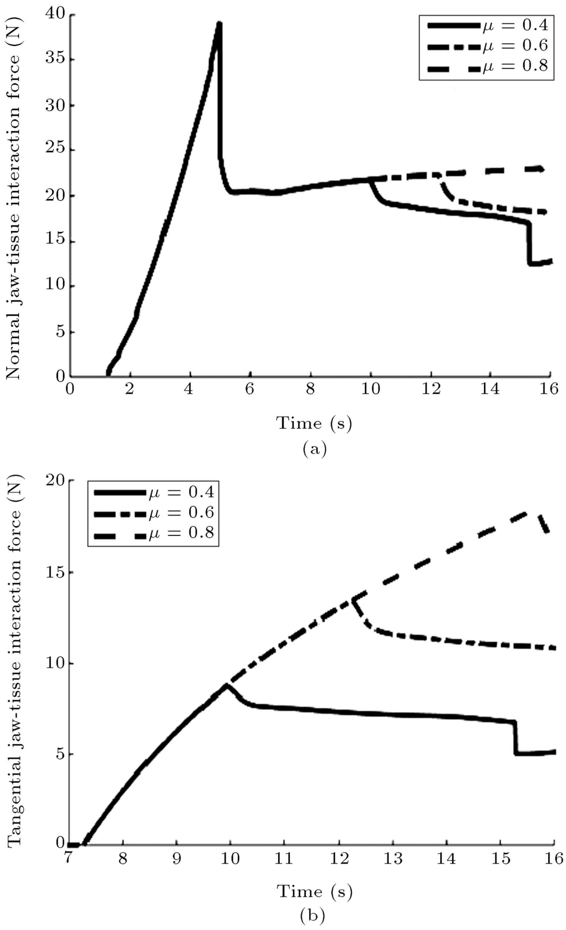


Figure 9. The normal (a) and tangential (b) force components of the tool-tissue interaction for different coefficients of friction of $\mu = 0.4$, $\mu = 0.6$, $\mu = 0.8$.

characteristics of the spleen, including its non-linear viscoelastic behavior, reasonably. The update rates for force computation and graphical rendering were 150 Hz and 25 Hz, respectively, on a conventional PC, for an MSD model with 236 nodes. The normal stress distribution in the lateral cross section of the spleen tissue at initial grasping is also illustrated in Figure 8, indicating high stresses under the grasper jaws. This stress was computed for each node as the normal component of the total inter-nodal force vector, acting on that node, divided by the area of the polygon formed between the mid points of its inter-nodal springs.

The tool-tissue interaction forces, produced between the spleen tissue and the jaws of the grasper, are illustrated in Figure 9. The normal component of the force interaction shows a nearly linear increase when the jaws are closing, and then a sharp decrease, when they are stopped; this is due to the stress relaxation behavior. After application of the pull force, a slight decrease is also observed in the normal interaction force. For the tangential force, the model predicted a nonlinear increase which was then followed by a decreasing pattern when the sliding started. The effect of the friction coefficient on the normal and tangential force components of the tool-tissue interaction are also illustrated in Figure 9. Higher friction did not affect the normal component of the grasping force, however, its decrease, due to slippage, started at a later time. The effect of the friction coefficient on the tangential force component was quite considerable. With higher frictions, the tangential force increased to larger magnitudes and the slippage appeared at higher pulling forces.

5. Discussion

Surgical simulation systems facilitate a safe and efficient training process by providing a virtual environment in which the trainee can repeat the surgical procedure unlimitedly at different situations [3,21]. A challenging requirement of these systems, however, is real-time performance, which highlights the importance of the biomechanical modeling approach. Moreover, not only the biological soft tissues often exhibit complicated mechanical behaviors, including nonlinearity and rate dependency [5,8,9], but also the modes of tool-tissue interactions are quite diverse during surgery. The simulation system needs to be able to simulate a wide variety of surgical tasks, e.g. indentation [5,12], needle insertion [13], cutting [22,23], and grasping [14-16]. As a result, it is much challenging to integrate an appropriate mechanical model of the tissue with the complicated and diverse tool-tissue interactions happening during surgery [5].

In general, the continuum based models provide a more accurate mechanical simulation of the organs' mechanical behavior. However, major limitations of such methods are the complexity of implementation [24] and their high computational cost. There have been attempts to improve the efficiency of FEM by using the condensation technique and preprocessing calculations [25]. However, a real time FEM based simulation is only achievable under major assumptions and simplifications. In particular, linear elasticity has often been used as a trade-off between biomechanical realism and real-time computation [24].

The discrete models, such as the mass-spring model used in this study, allow for large deformations and displacements, and are fairly easy to implement [5]. Despite the lower accuracy, their simplicity of implementation and their relatively low computational cost make them more suitable for surgical training simulators. This would be more critical for large soft organs for which the biomechanical behavior is highly complicated. A simulation of large intra-abdominal organs behavior not only needs a non-linear hyper elastic model, due to their large deformations, but also a viscoelastic time-dependent formulation [15,16].

The MSD modeling approach employed in this study was capable of simulating the complicated mechanical behavior of nonlinear viscoelastic objects in real time with a reasonable accuracy. The predictions of the model for the deformations of the synthetic test sample during three fingered grasping were in good agreement with the experimental results (Figures 6 and 7). The resulting geometries and curvatures were qualitatively similar under the jaws and in non-contact areas (Figure 6). Also, the force-deformation behaviors were quantitatively close, particularly when grasping force was not very large (Figure 7). The step-like pattern of the force-deformation curve, predicted by the model, was due to the node-based approach we used in this study for simulation of the contact between grasper jaws and tissue. With the grasper closing, new nodes came into contact with the grasper jaws, each causing a sudden increase in the test sample's stiffness. This behavior can be improved if a larger number of mass nodes are considered within the model, particularly at its contact area with the grasper.

For grasping of the spleen organ, there was no experimental basis to be used for verification of the model's prediction. However, the general characteristics of the tissue's complicated biomechanical behavior under different tool-tissue force interactions were well exhibited by the model. For instance, the tissue exhibited a stress relaxation behavior when the deformation was kept fixed, i.e. the jaws stopped closing (Figure 9). These results suggest that the model is well capable of simulating the complicated viscoelastic behavior of the spleen organ. Moreover, the model could reasonably replicate the two modes of slippage-free and slippage-included grasping, under low and high pull forces, respectively. Finally, the predictions of the model for the effects of the friction coefficient on the grasping behavior (Figure 9) were reasonable and provided further evidence for the model's capability to be used for simulation of the interactions of the surgical grasper and a large soft organ. Nevertheless, the fact that the model's predictions for the spleen organ were not validated against the appropriate experimental data is an important limitation of the present study, which was caused due to the technical difficulties involved in the experimental tests of biological tissues.

As a conclusion, it might be suggested that the proposed MSD modeling approach could effectively fulfill the basic requirements of surgical simulation systems, i.e. a real-time and generally realistic response. Although the accuracy for representation of the biomechanical behavior of the biological soft tissues might not be as high as that of the continuum based approaches, e.g. finite element method, it is adequate for a training system in which a sensible response is sufficient. The proposed MSD modeling approach provides a reasonable trade-off between the biomechanical realism and real-time computation requirements of a surgical training simulation system.

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