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Neurosurgery Simulation Using Non-linear Finite Element Modeling and Haptic Interaction

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Abstract

Real-time surgical simulation is becoming an important component of surgical training. To meet the real-time requirement, however, the accuracy of the biomechanical modeling of soft tissue is often compromised due to computing resource constraints. Furthermore, haptic integration presents an additional challenge with its requirement for a high update rate. As a result, most real-time surgical simulation systems employ a linear elasticity model, simplified numerical methods such as the boundary element method or spring-particle systems, and coarse volumetric meshes. However, these systems are not clinically realistic. We present here an ongoing work aimed at developing an efficient and physically realistic neurosurgery simulator using a non-linear finite element method (FEM) with haptic interaction. Real-time finite element analysis is achieved by utilizing the total Lagrangian explicit dynamic (TLED) formulation and GPU acceleration of per-node and per-element operations. We employ a virtual coupling method for separating deformable body simulation and collision detection from haptic rendering, which needs to be updated at a much higher rate than the visual simulation. The system provides accurate biomechanical modeling of soft tissue while retaining a real-time performance with haptic interaction. However, our experiments showed that the stability of the simulator depends heavily on the material property of the tissue and the speed of colliding objects. Hence, additional efforts including dynamic relaxation are required to improve the stability of the system.

Keywords

surgical simulation; non-linear biomechanics; finite element method; haptic rendering

1.PURPOSE

Recent studies have shown that surgical simulators can enhance a standard surgical training, which typically only involves observations of a neurosurgery, and therefore improve the quality of health care.¹ With the current trend of limited hours available for surgical training,² the importance of the simulation could become increasingly significant. Physically-based modeling techniques provide accurate simulation of deformable organs^{3, 4} and have been applied to surgical simulations⁵⁻¹⁰ and image-guided surgery.^{11, 12} Among the physical models and numerical methods applied to medical simulations, non-linear

elasticity models¹³ are especially of interest because of improved accuracy in modeling human organs enduring large deformations.^{3, 4}

However, due to the real-time performance requirement, the accuracy of surgical simulators is often sacrificed in terms of the physical model and the numerical method. Simpler models such as linear mass-spring,⁶ boundary element modeling,⁹ and co-rotational (linear) FEM¹⁴ have been applied to real-time simulations with haptic interactions. Recently, non-linear FEM using an explicit integration scheme has been applied to real-time applications with the help of recent GPUs,^{5, 7, 8} but has not been integrated with more sophisticated contact models and haptic rendering, which run at much higher rates than most real-time applications. We aim to overcome these limits for neurosurgery simulation by adopting recent GPU technology and virtual coupling, so that the non-linear FEM can be integrated with haptic interactions.

2. METHODS

We employ the total Lagrangian FEM with an explicit time integration for the neo-Hookean elasticity model.¹³ The 3D domain (deformable object) is subdivided into tetrahedra (*elements*), each consisting of four vertices (*nodes*). The partial differential equation (PDE) defined over the domain is discretized in space using piecewise-linear *shape functions* defined with the node and element topology. We then numerically integrate with respect to time to compute the displacement vectors at the nodes. The *total Lagrangian* formulation of the PDE uses spatial derivatives of nodal displacements with respect to the initial (undeformed) configuration of the domain, and therefore eliminates the need to recompute the mass matrix and the spatial derivatives of shape functions in each time step. The explicit time integration avoids solving non-linear equations in each time step and simplifies the computation of contact response.

For an explicit integration method to remain stable, the time step needs to be smaller than the *critical time step* given by the CFL condition: $\Delta t_{crit} = L/c \propto L/\sqrt{E}$, where L is the minimum characteristic size of elements, c is the wave propagation speed of the material, and E is the stiffness value (*Young's modulus*). Smaller time steps deteriorate the performance of the simulator, but the problem is alleviated due to the nature of the application: the brain tissue is soft, and sudden movements are seldom seen in a neurosurgery, therefore mesh distortion is limited. Volumetric mesh generation is carefully controlled to avoid small tetrahedrons. In addition, we employ the dynamic relaxation method¹⁵ to add damping into the system in order to further improve the stability. The per-element and per-node operations in the FEM are parallelized on the GPU to achieve real-time performance.^{5, 7}

The contact response is modeled using the constraint-based forces (apply just enough forces to separate the inter-penetrating objects)¹⁴ applied after each time step. On the other hand, haptic rendering (computation of force feedback) must run at a much higher rate than the simulation, the collision detection, and the contact handling. In a scene involving interactions between a deformable object and an “instrument” controlled by the haptic device, we use a virtual coupling¹⁶ to decouple the haptic rendering from the simulator. The

actual configuration (location and orientation) of the haptic device instantly maps to a virtual object that is not simulated in the surgical simulation. In another system, the virtual object is connected to the instrument with linear springs, which generate the feedback forces. The interaction between the instrument and the deformable object is handled in the ordinary simulation loop, while the spring connections between the virtual object and the instrument are simulated in a different thread that runs at a higher rate to generate the force feedback. Namely, we have two simulation loops running in separate threads: one for the scene including the deformable object and one for the haptic rendering. The instrument is the only object that is affected by both loops.

3. RESULTS

We use the library CHAI 3D¹⁷ to handle a PHANTOM Omni haptic device (SensAble, MA, USA) and the real-time visualization. The tetrahedral mesh for the FEM is generated with TetGen.¹⁸ Per-element and per-node computations in the FEM are implemented for the GPU (NVIDIA Quadro 2000) using the Compute Unified Device Architecture (CUDA) SDK.¹⁹ We currently use three threads in the system: one for deformable body simulation and contact response, one for haptic rendering, and one for visualization.

Our current system handles poking of the soft tissue modeled as a neo-Hookean solid. Figure 1 shows screenshots of real-time simulations of a cube and a brain with haptic interaction. The cube model consists of 937 nodes and 3543 tetrahedral elements, and the brain model consists of 529 nodes and 1943 tetrahedral elements. The time step is 0.001 second in both simulations.

Figure 2 shows the plots of critical time step versus Young's modulus in the sphere-cube scene (shown in the first row of Figure 1), with the sphere moving at different speeds for a fixed distance. When the speed of the sphere is small (1 m/s), the plot follows the $\Delta t_{crit} \propto 1/\sqrt{E}$ relation. However, when the speed is higher, t_{crit} depends more on other factors such as mesh distortion, which affects L . Softer materials can be distorted easier and therefore can be less stable than stiffer materials under a certain scenario of interactions. Our experiments show that with higher contact speeds, t_{crit} actually increases with E when E is small, and above a certain threshold value of stiffness, t_{crit} again decreases with $\sqrt{E}$. The threshold values we found are 16 kPa and 30 kPa when the sphere moves at 3 m/s and 5 m/s, respectively, which indicates that the effect of distortion is more prominent when the contact speed is higher. The results indicate that our system is most suitable for applications with soft tissues and slow movement of colliding objects, which is often the case in neurosurgery simulations.

4. CONCLUSIONS

We presented on-going work on real-time neurosurgery simulation with haptic interface. We overcome the performance issue of non-linear FEM with the total Lagrangian explicit formulation and GPU acceleration, and the haptic rendering is handled with a virtual coupling method. The system provides accurate biomechanical modeling of soft tissue while retaining a real-time performance with haptic interaction. However, our experiments showed

that the stability of the simulator depends heavily on the material property of the tissue and the speed of colliding objects. Hence, additional efforts including dynamic relaxation are required to improve the stability of the system. Furthermore, we will integrate our simulator with a seven degree-of-freedom haptic device and explore tissue pinching and cutting, which require fast re-meshing algorithms.

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Figure 1. Screenshots of the interactive simulation of a cube and a brain; left: rest state; right: after interaction with a rigid ball. The orange ball represents the haptic input.

Figure 2.

Plots of critical time step (sec) versus Young's modulus (kPa) in a simulation where a rigid sphere collides with a deformable cube (see first row of Fig. 1); each curve shows the results when the sphere moves in a different speed (m/s). The stability depends on both the stiffness and the speed of colliding object.