Artery Soft-Tissue Modelling for Stent Implant Training System

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ABSTRACT

Virtual reality technology can be utilised to provide new systematic training methods for surgical procedures. Our aim is to build a simulator that allows medical students to practice the coronary stent implant procedure and avoids exposing patients to risks.
The designed simulation system consists of a virtual environment and a haptic interface, in order to provide both the visualization of the coronary arteries and the tactile and force feedback generated during the interactions of the surgical instruments in the virtual environment.
Since the arteries are soft tissues, their shape may change during an operation; for this reason physical modelling of the organs is necessary to render their behaviour under the influence of surgeon's instruments. The idea is to define a model that computes the displacement of the tissue versus time; from the displacement it is possible to calculate the response of the tissue to the surgical tool external stimuli.
Information about tools displacements and tissue responses are also used to graphically model the artery wall and virtual surgical instrument deformations generated as a consequence of their coming into contact.
In order to obtain a realistic simulation, the Finite Element Method has been used to model the soft tissues of the artery, using linear elasticity to reduce computational time and speed up interaction rates.

Keywords: Surgery Simulation, Stent Implant, Soft Tissue Modelling, Haptic Interface, FEM.

1. INTRODUCTION

The main trend in modern medicine development is the use of therapy such as endoscopy and laparoscopic surgery. These methods allow the diseased area to be reached by means of a small incision in the body. During the operation a monitor shows what is happening inside the body, while touch sensation comes via the long instrument. There is a big difference compared with the open surgery, where the surgeons have full visual and manual access to the organ. To learn the correct steps of this closed procedure, the surgeon has to learn to acquire hand-eye coordination and manual dexterity.
For this reason minimally invasive surgical procedures require a new training method that is very different from traditional techniques; frequent practice should be performed in a safe environment which mimics the anatomy and physiology of the body as closely as possible to ensure adequate transfer of skills [1], [2].
Virtual reality technology with innovative haptic interfaces can utilised to provide new systematic training methods for surgical procedures. In order to transfer surgical skills it is necessary to provide a realistic interactive environment that reflects the real life procedure.
Surgical simulation and training systems demand both real-time and physically realistic modelling of deformable tissues.
The realism of the interactions can be enhanced using a haptic interface provides force feedback sensations which are as close as possible to reality, but computations performed in real-time are required for this.
Only by obtaining a realistic representation of the anatomy, a real-time tissue-tools interaction and a fair force feedback provided with a haptic device, can the training experience be transferred to actual patient care.
The utilisation of surgical training devices combined with a computer simulation means avoiding the problems of in vivo practice or the use of animal models. The advantage of using a virtual reality simulator is that it provides a safe controlled medical environment to practise and to prepare procedures; different scenarios can be built into the simulator to provide a wide experience that can be repeated as needed.

2. DEFORMABLE TISSUE MODELS

To obtain a correct representation of deformability it is necessary to compare the computed soft-tissue model to the actually deformed tissues. Biomechanics has proposed complex mathematical models for representing the deformation of soft tissues and computer graphics has developed many algorithms for the real-time computation of deformable objects.
Instead of using implicit nodal methods, that require the solution of matrix systems, explicit models (e.g. mass-spring models) have frequently been used because they are very easy to implement and they yield reasonable speeds.
Mass-spring mesh is a set of point masses connected by elastic links; a mass is assigned to each node in addition to a damping coefficient. Springs exert forces on neighbouring points when a mass is displaced from its rest positions and the spring behaviour is governed by a deformation law (typically Hooke's law). The amount of stiffness of the springs can be derived from the intensity of voxels in a CT-scan image; in this way the stiffness is proportional to tissue density and therefore to the Hounsfield units. The mass-spring method doesn't require a continuous parameterization and can be used to model cutting or suturing simply by removing or adding connections between vertices. The behaviour of a mass-spring mesh depends heavily on its topological and geometric configuration. In addition, configurations with large forces (e.g. nearly rigid objects) lead to stiff differential equations with a poor numerical stability, requiring small time-steps of the integration. Nevertheless a mass-spring model is an easily understandable concept, which is simple to implement and has low computational demands.
Therefore, mass-spring models are used in a wide range of computer graphics and virtual reality applications, e.g. in the animation of facial expressions, the simulation of cloth motion and the modelling of inner organs in surgery simulations.

Several improvements to spring models have been proposed, specifically with regard to their dynamical behaviour.
Cover et al. [3] were the first to present a real time model for gall bladder surgery simulation.
Kühnapfel et al. [4] used a mass-spring model to simulate a realistic interaction between surgical tools and organs in the KISMET system, a virtual reality training system for minimally invasive surgery. Virtual organ geometry is modelled as elastic body and the control points of the object surface form together with additional internal nodes an elastic 3D-mesh of virtual mass points, which are interconnected by virtual springs with damping elements.
Gibson [5] proposed a "ChainMail" model, where volume elements are linked to their nearest neighbours. Each node must satisfy a given maximum and minimum distance constraint to its adjacent nodes. When an element is moved and one of its constraints is violated, a displacement of the respective neighbouring element takes place. In this way small displacements of a selected point in a relatively slack system result in only local deformations of the system, while displacements in a system that is already stretched or compressed to its limit cause the whole system to move.
Brown et al. [6] present an algorithm for animating deformable objects in real-time and the target of the application domain is the microsurgery. They have designed an integrated system for simulating and suturing of small blood vessels.
Mass-spring methods approximate the object as a finite mesh of points and the equilibrium equation at the mesh points is computed in discrete way; more accurate physical models treat deformable objects as a continuum.
In the Finite Element Method (FEM) a problem is stated in a continuous way, but solved for each basic element (triangles, quadrilaterals, tetrahedral, etc) that describe the object's shape. Simple interpolation functions within these elements make the problem numerically tractable; appropriate boundary conditions guarantee a physically correct solution.
One major advantage of the FEM is the scalability of the solution method because with the same mesh structure it is possible to increase or decrease the precision and complexity of the model, obtaining an advanced model or a less advanced but faster model.
The FEM may be more accurate than the mass-spring model, but it is more computationally expensive. In medical simulations FEM may address better the needs of applications like the pre-operative surgical planning where the precision is prior to a real-time response and the computations can be done off-line; in this case it provides more accurate patient-specific results.
In real-time simulation, several modifications and simplifications must be introduced to reduce computational time in order to find a trade-off between biomedical realism and real-time interaction.
Bro-Nielsen et al. [7] use an approach, called "Condensation", to achieve real-time performance for the application of 3D solid volumetric FEM. This improvement involves compressing the linear matrix system resulting from the volumetric FE model to a system with the same complexity as a FE surface model of the same object.
The cutting or suturing procedures require a redefinition of the finite element model and, for this reason, they can not be applied as the calculation cost is too high.
Cotin et al [8] propose a new method, called "tensor-mass model", that is as simple to implement and as efficient as spring-mass models, but is based on continuum mechanics and linear elasticity theory. In this method the stiffness matrix only depends on the material characteristics within a tetrahedron and, for this reason, it is possible to simulate the cutting and tearing of soft tissue. A limited number of elements (around one thousand) is allowed for a real-time situation. Moreover in [8] a "hybrid elastic method" is proposed making it possible to cut and deform large anatomical structures. A simulation of a hepatectomy has been chosen to demonstrate the approach efficiency.

3. SIMULATOR REQUIREMENTS

The simulation combines a visualization system, where a representation of the real environment is reproduced, and a haptic device which provides a force feedback to the user during the interaction in the virtual environment.
The information flow forms a closed loop: the motion of the medical instruments is reproduced in the virtual environment, the contact forces are computed and the loop is closed by generating these forces on the surgeon's hand by means of the haptic interface.
To obtain a realistic simulation the main difficulty is related to the real-time constraint that imposes a very high frequency in forces computing. The realism of the sensation of force is highly dependent on the physical realism of the model. The modelling of the virtual environment involves geometrical description of the organs, their behaviour and their interactions with the medical instruments. Since the arteries are soft tissues, their shape may change during an operation; for this reason physical modelling of the organs is necessary to render their behaviour under the influence of surgeon's instruments.
The physical modelling gives the constitutive equations that describe the mechanical properties of the real body.
The two main constraints for the modelling of soft tissues are deformation accuracy and computational time. For surgery procedure training, computational time is more important than the accuracy of deformation in order to achieve smooth user interaction. The desired realism of the physical models must be balanced against the need for speed.
The real time deformation of soft tissue is an important constraint for medical simulator. Delay between the user action and the environment reaction disturbs the perception of correct interaction in the virtual environment and reduces the immersion sensation.
There are several contributing causes to latency: communication between the haptic interface and the virtual environment, the computation time for collision detection, force feedback and deformation; the total latency is not the sum of those delays because many elements are asynchronous. The latency depends greatly on hardware used and it is important to reduce them to their minimum value.
The ability of the operator to learn from a computer-simulated system is directly connected to the bandwidth of the system; an acceptable bandwidth for visual feedback is in the range of 20-60 Hz, while an acceptable bandwidth for haptic feedback is in the range of 300-1000 Hz.

4. CORONARY STENT IMPLANT PROCEDURE

A coronary stent implant is a therapeutic cardiac procedure that is used in association with balloon angioplasty to open up a blocked coronary artery. After implantation, the coronary stent

becomes a permanent implant to hold the artery open and prevent it from closing back down.
The coronary stent is a small, slotted stainless steel tube that will be passed through the catheter in the femoral artery. The doctor initially advances a long, thin guide wire through the sheaths and up across the blockage in the coronary artery. This initial widening of the blockage is necessary to pass the stent catheter through the blockage. Next, the stent catheter is threaded onto the guide wire. The stent is placed around a deflated balloon.
The stent catheter is positioned in the centre of the blockage and the balloon is slowly inflated and the stent expands, spreading the blockage apart. When the balloon is deflated, the stent remains expanded. The stent presses slightly into the wall of the coronary artery, keeping it open (fig.1). The balloon catheter and guide wire are then removed and the stent holds the artery open reducing the rate of restenosis.
This intervention requires perfect knowledge of the three-dimensional structures of the vessels, hand-eye coordination and manual dexterity in order to perform the implantation safely.
Our aim is to build a tool that allows medical students to practice this delicate operation avoiding exposing patients to risks.

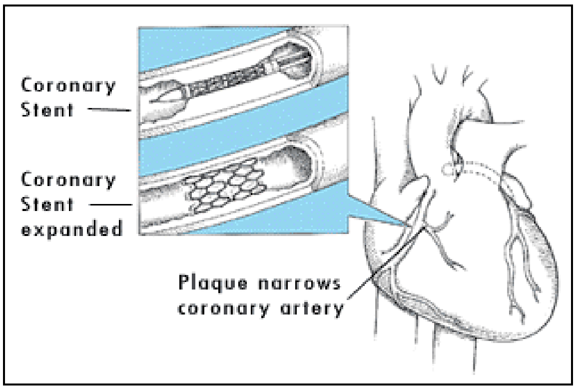

fig.1: Coronary stent implant procedure

5. TRAINING SYSTEM DESCRIPTION

The designed training system consists of a virtual environment and a haptic interface, in order to provide both the visualization of the coronary arteries and the tactile and force feedback generated during the interactions of the surgical instruments in the virtual environment.
The user interacts with the training system using the haptic interface. The haptic interface reproduces the real surgical instruments that the cardiologist manipulates during the stent implant operation. Data acquired from haptic device sensors are used to represent the instrument position in the virtual environment and to determine possible collisions between virtual objects.
Movements of haptic devices lead to changes in the virtual environment representation; collisions between virtual objects produce forces that are replicated on the surgeon's hand by the haptic interface.
The scheme of the simulation system is shown in fig.2.
In particular:

- the haptic interface is the electromechanical component responsible for reading the user's position and providing him with the appropriate force components;
- the collision detection and response modules are in charge of detecting any collision in the virtual scene and of computing the force that will be exerted on the user by means of the haptic device;
- the environment animation and description module are in charge of mathematically modelling the virtual objects behaviour;
- the physical modelling module is in charge of describing the mechanical properties of the organs.

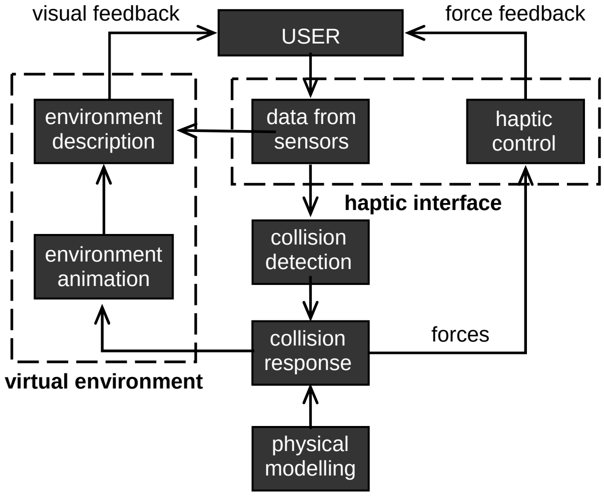

fig.2: Scheme of the simulator

The haptic interface, designed and built by the PERCRO Laboratory of Scuola Superiore S. Anna, Pisa, Italy, reproduces the real surgical instruments used by the cardiologist. This device is provided with two degrees of freedom that produce force and torque resistance. In particular, the system responds to the following forces applied by the user:

- the longitudinal forces in the form of push and pull movements;
- the torque forces in the form of twisting around the longitudinal axis.

The scheme of the haptic interface is shown in fig.3.

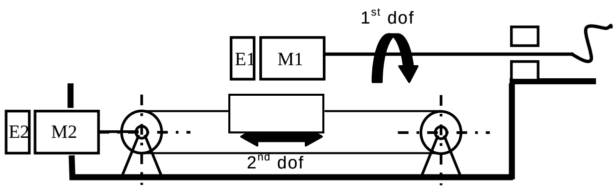

fig.3: Scheme of the haptic interface

The virtual artery model has been built using advanced CAD modelling techniques and the anatomical models described in medical literature.
The system should guarantee the accuracy of the simulation and real time interactivity, allowing medical students to practice this operation as it would be in the real world.

6. PHYSICAL MODELLING

To obtain a realistic interaction it is necessary to provide a force feedback taking into account the physical behaviour of the organs. Moreover, delay between the user action and the environment reaction reduces the immersion sensation and, for this reason, it is necessary to have real-time interaction. Therefore, the physical modelling module is used both to provide information about the artery wall deformations (used for the rendering of the virtual scene) and to calculate the haptic sensation to the surgical tool stimuli.

The application of FEM in a real-time system has proven difficult because the force vectors and the mass and stiffness matrices must be re-evaluated as the object deforms. This re-evaluation is very costly in terms of computational time.

In our system the “Quasi-Static Pre-Computed Linear Elastic Model” has been used, in order to enhance the realism of the model reducing the computational time required by the standard FEM approaches [9].

Since the deformations of the artery wall as a response to mechanical stress acting on it are local to the contact area, it is reasonable to consider them linear elastic. Furthermore, it is possible to prevent large deformations of the artery, because the force computed by the application and replicated on the physician’s hand increases proportionally to the deformation.

In addition, the velocity of the medical instruments, and thus of the artery mesh nodes in case of contact between these objects, is small enough for the mesh to achieve static equilibrium at each instant. This is a reasonable assumption in a catheterization procedure where the medical instruments have a slow motion.

Finally, the artery is an empty body. Studies in biomechanics have shown that the blood flow in the vessels can be considered continuum and incompressible. So, the artery becomes a solid body and can be modelled as a volumetric deformable object, in which it is possible to move.

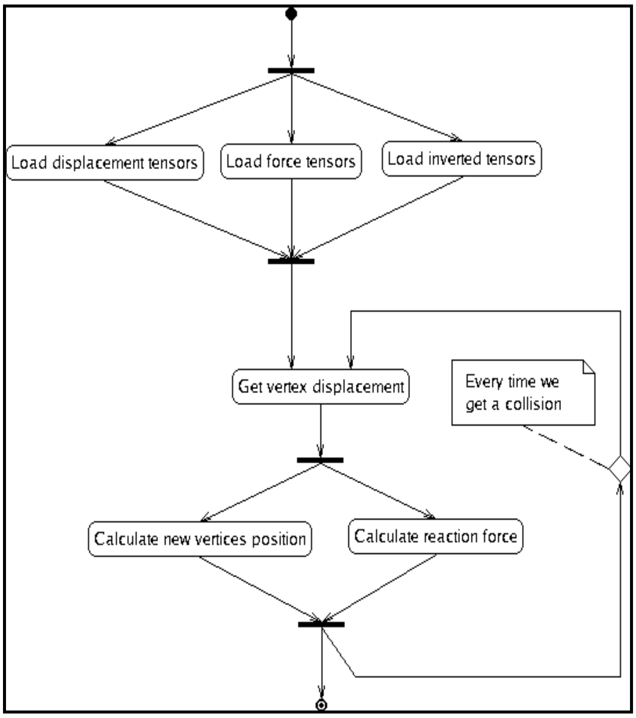

fig. 4: Activity diagram of the real-time phase

Under these hypotheses, we can take advantage of the linearity and the superposition principle, exploited by the Quasi-Static Method, to model the artery wall deformations.

Thus, the artery is represented by a tetrahedral mesh and the solution of the linear system $[K]\cdot u = f$ is calculated; [K] is the stiffness matrix, u is the displacement of all nodes and f represent the external forces.

The size of matrix [K] is (3N x 3N), where N is the number of nodes. For this reason the computational complexity is a linear function of the mesh size (number of mesh nodes) and consequently the possibility of simplifying the geometric representation is very important.

Interactive rates of deformation are obtained in a two-step process:

- a pre-processing stage, performed off-line and used to compute a set of elementary deformations of the model;
- a real-time stage, where all deformations are computed as a linear combination of the previous pre-computed ones.

To perform the pre-processing step it is necessary first to specify a set of mesh nodes that remain fixed during the interaction in order to guarantee deformation and not translation.

After this, for each free node on the surface of the mesh, an elementary displacement is set and some quantities (displacement tensor and force tensor) are computed. These quantities are used in the real-time stage for the computation of the displacement and force associated to a node.

fig. 4 shows the activity diagram of the real-time phase.

The pre-processing stage can take few minutes or several hours depending on the model size and the desired accuracy level.

The result of the pre-processing stage, saved for further simulations, needs to be obtained only once for a given model.

	Test 1	Test 2	Test 3	Test 4	Test 5
tetrahedrons	175	262	533	989	2506
nodes on surface	60	68	84	131	249
triangles	104	120	152	246	474

Table 1: Test specifications

The described algorithm has been tested on a personal computer with a Pentium IV 1 GHz CPU, 512 MB RAM and Linux Mandrake 9.1 operative system.

In order to verify the times to carry out the off-line and the real-time elaborations in relation to the complexity of the model and the desired accuracy level, five tests have been performed increasing the number of nodes.

Table *1* shows the test specifications.

fig. 5 and fig. 6 respectively show the off-line and the real-time elaboration times.

As expected, the off-line computational time increases steeply as the number of nodes does, while the time required for the real-time elaboration remains sufficiently low and it allows interactions to be performed without a significant delay.

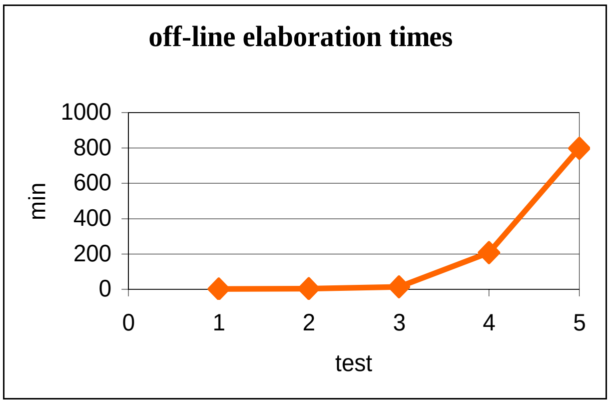

fig. 5: Off-line elaboration times

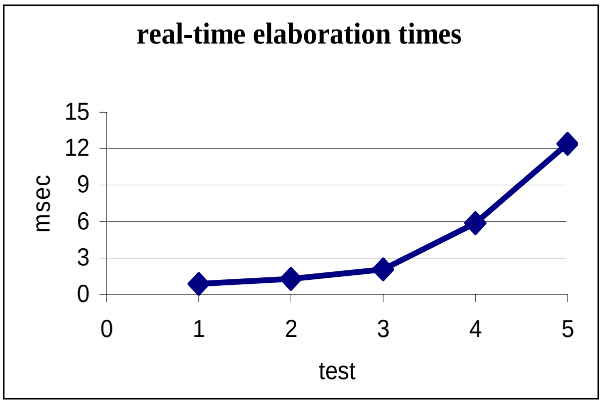

fig. 6: Real-time elaboration times

7. CONCLUSIONS AND FUTURE WORK

This work represents the first step in realizing a realist simulation system to perform training on the coronary stent implant procedure.

In order to enhance the realism of the simulation, we modelled the deformations of the soft artery wall using the FEM. In particular, to obtain a good trade-off between biomechanical realism and real-time computation we exploited the Quasi-Static Method.

Tests have been performed to estimate the computational time of the algorithm executed on input of growing complexity.

To validate the system and to guarantee the accuracy of the simulation, a test phase in collaboration with cardiologists will be carried out.

Further enhancements will be done in terms of visual rendering and haptic sensations.

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