医用画像処理に基づく内視鏡下手術シミュレーション

Endoscopic Surgery Simulation Based on Medical Image Processing

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Abstract
This paper explains about endoscopic surgery simulation based on medical image processing. Due to the progress of medical imaging devices, it is possible to acquire very precise volumetric images of a patient. These precise volumetric images enable us to simulate endoscopic surgery. This paper classifies types of surgical simulation into three categories: (a) simulation based on simple visualization, (b) deformation simulation of single organ, and (c) integrated simulation of endoscopic surgery. Here, we show examples of surgical simulation including virtualized endoscopy system, TBLB path simulation, virtual unfolded view generation of the stomach, and virtual laparoscopic image generation. Several simulation images are also presented here.

Key words
Virtual endoscopy, Endoscopic surgery, Surgical simulation, Deformation Virtual laparoscopy,

1．はじめに

近年の医用イメージング装置の発展により高精細な人体３次元像を取得することが可能となった．例えば，マルチスライス型 CT 装置を用いればわずか数十秒の間に人体体幹部の３次元 CT 像を取得することができる．ここで撮影可能な画像は一辺がおよそ 0.5mm 程度のボクセルからなる等方解像度画像である．このような高精細人体３次元画像を用いることで，種々の手術治療シミュレーションを行うことが可能となっている．特に医用画像処理技術により種々の臓器・血管領域等を抽出し，これらの臓器領域に外からの力による変形を計算し，その結果を可視化することで手術治療支援のシミュレーションを行うことが可能となる．患者個々のデータ（医用画像）に基づきこれらのシミュレーションを実行可能であり，手術計画立案に極めて重要であるだけでなく，教育目的のシミュレータとしても利用できる．本教育講演では，特に内視鏡手術治療のシミュレーションに着目し，人体内部を内視鏡で観察可能な仮想化内視鏡から腹腔鏡手術をシミュレーション可能な手法まで，医用画像処理技術の観点から解説したい．

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2．医用画像処理に基づく内視鏡下手術シミュレーション

ここでは，医用画像処理に基づく内視鏡手術シ

ミュレーション手法に関して，簡単にまとめる．特に単に「シミュレーション」と書く場合，様々なレベルでの「シミュレーション」が含まれるためため，組織的に整理しておくことは重要であると考える．

「医用画像を用いた手術治療シミュレーションシステム」を利用する場合の一番の目的は，「患者ごとに最適な手術治療法をシミュレーション画像を通じて検討する」ことであろう．例えば，患者ごとの解剖学的構造情報を把握するためにシミュレーション画像を生成し，最適な手術治療法をこの画像を観察しながら計画することが挙げられる．ここでは，単純に各患者の解剖学的情報を表示することにより腹腔鏡下手術をシミュレーションすることなどが行われる．単なる解剖学的な構造情報表示であるが，これも「手術シミュレーション」と記される．もちろん，腹腔鏡下手術で行われる手技等をコンピュータ上で正確に再現し，必要であれば臓器変形等をシミュレーションするシステムも「シミュレーション」である．そこで，「医用画像を用いた手術シミュレーション」システムを考えてみると、

(1) 単純可視化シミュレーション型
(2) 単臓器変形シミュレーション型
(3) 統合型手術治療シミュレーション型

の３つの型に分類できる．これらはさらに，

(a) 患者毎モデル型
(b) 一般モデル型

の２つに分類できるが，先述のように本稿では，(a)の患者毎モデル型を主として取り扱う．(1)のタイプのシミュレーションは 3 次元医用画像を基にして術野に類似した映像を生成するものである．臨床の場ではこれも手術シミュレーションと呼ぶ．例えば後述する仮想化内視鏡システムは[1)-7)]，３次元医用画像を基にあたかも内視鏡で観察したかのような映像をシミュレートすることができ，(1)のタイプに属すといえる（仮想化内視鏡システム発表初期のころは内視鏡のシミュレーションと呼ばれていた）．また，患者腹腔内の血管走行情報を表示し，リンパ節等を重畳表示した画像もこのタイプに分類することができる．(2)ならびに(3)は，臓器変形計算を伴った手術シミュレーションであり，狭義での手術治療シミュレーションと言える．(2)のタイプは，例えば肝臓といった個々の臓器の変形をシミュレーションするものである．例えば，肝臓を切開するシミュレーションなどがこのタイプの手術シミュレーションに該当する[8)-10)]．あるいは，後述の胃の仮想展開像生成などもこの型の手術シミュレーション手法といえよう．(3)のタイプは，腹腔鏡下手術などを総合してシミュレーションするものである．現在のところいくつかの商用機が登場しているが[11)]，あらかじめ用意されているコンテンツを用いて手術シミュレーションを行うものがほとんどである．

３．医用画像処理に基づく内視鏡下手術シミュレーション

３.１ 単純可視化シミュレーション型

1) 概説

この分類のシミュレーション手法は，入力画像が与えられたときに，できる限り術野に近い映像を生成するものである．あるいは，血管の走行情報とリンパ節位置関係など手術計画立案に必要な情報を，解剖学的構造を可視化することによって提示するものもこの分類に含まれる．このようなシミュレーション手法における技術的事項としては，(1) 臓器領域（実質臓器・脈管系等）セグメンテーション，(2) 可視化，(3) パス生成，などが含まれる．

2) 仮想化内視鏡システム

ここでは，医用画像処理を用いた最も基本的な内視鏡下手術シミュレーションシステムとして，仮想化内視鏡システムについて解説する[1)-7)]．これは先の分類の(1)単純可視化型のシミュレーションに相当する．仮想化内視鏡システムの初期の報告は 1993 年頃になされているが．基本的な技術事項を解説することは極めて有用と考える．

仮想化内視鏡システムとは３次元 CT 像．3 次元 MRI 画像といった３次元医用画像を基にあたかも患者体内を内視鏡で観察したかのような画像を得ることのできるシステムである．先述のように，仮想化内視鏡システムに関する初期の報告は筆者の研究グループも含めた複数の研究グループから報告されている．現在では，ほとんどの医用画像処理ワークステーションに仮想化内視

鏡モードが搭載されている．仮想化内視鏡システムを実現する上での技術的ポイントを列挙すると，(1) 透視投影，(2) 視点を臓器内部に設定，(3) 対話的描画，(4) フライスルー，などがある．(1)の透視投影では，シミュレーションが対象とする内視鏡のカメラ内部パラメータ（画角等）に応じた透視投影パラメータが設定される．(2)は内視鏡画像を生成するのであるから，何らかの方法で対象とする臓器内部に視点位置を設定しなければならない．先述の透視投影の投影中心が臓器内部に設定されることになる．(3)の対話的描画法では，高速に仮想化内視鏡画像を生成するために必須である．一般的にはサーフェスレンダリング法，ボリュームレンダリング法[12)]の２つが用いられる．サーフェスレンダリング法では，臓器表面形状を三角形パッチの集合で表現し，それをグラフィックスボードにより高速に描画する．ボリュームレンダリング法では，原画像の濃度値に対する色・不透明を設定することで（伝達関数と呼ばれる），原画像から直接的に臓器形状を描画する．ボリュームレンダリング法では，視点位置から投影面の１点を通るレイを考え，その上にサンプリング点(標本点)を設定する．各標本点における濃度値を補間により求め，その値を用いて伝達関数を参照することで色・不透明度を得る．その後，光源の方向と標本点でのグラディエントベクトルの方向を用いて陰影付けを行る．この処理をレイに沿って不透明度に基づき積算することで，投影面上１点での色を求める．この処理を投影面上全点に対して実行することで，３次元画像を得る．ボリュームレンダリング法はソフトウエアで高速に処理する手法[13)]のほかに，グラフィックボードが備えるプログラマブルシェーダーを利用した高速描画手法，専用グラッフィクスボードを用いた描画方法などが開発されている．(4)のフライスルー機能では，マウスなどの操作による臓器内部を自由なフライスルー，連続的な視点位置・視線方向の変更，マウスカーソルの指し示す方向への前進，視線方向を軸としたひねり，これまでの経路のトレースバックなどの機能が用意されている．これらの機能により３次元 CT 像．あるいは，３次元 MRI 像を基に内視鏡のシミュレーションを行うことができ，内視鏡下手術における術野画像に類似した映像を生成可能となる(**Fig. 1**)．

3) 仮想化内視鏡システムを用いた経気管支肺生検シミュレーション

仮想化内視鏡システムを用いることで，肺がんなどの異常が CT 像上等で発見された場合，気管支鏡を用いて生検を行う際のシミュレーションが可能となる．この場合，病変に最も近い気管支への経路を求め，その経路を仮想化内視鏡システム上で確認することになる．経路は通過すべき気管支の枝名（解剖学的名称）の集合として出力され，その経路に沿った自動フライスルーが行われる．気管支壁下に存在する解剖学的構造物も提示することが可能である，このような画面を確認することで，経気管支肺生検のシミュレーションを行うことができる(**Fig. 2**)．ここで経路を生成する場合には，病変部に最も近い気管支，病変が存在する肺葉内での経路などをコンピュータが自動的に求めることでシミュレーションが可能となっている．

4) 血管・リンパ節位置表示による手術シミュレーション

入力される原画像から画像処理技術により血管・リンパ節領域を取り出し，その情報を仮想化内視鏡システム上において表示することで，あらかじめ術野を確認するシミュレーションが可能となる．ここでも積極的な臓器の変形は行わないが，外科手術において需要となる動脈系・リンパ節を仮想化内視鏡画像上に表示することで，切除範囲を決定するといったシミュレーションを実行することができる．この際，外科手術で関心のある動脈系（例えば下腸間動脈など）をユーザが選択的に表示もしくは非表示とすることで，実際の切除を模したシミュレーションが可能である．この場合，血管・リンパ節の抽出が問題となるが，前者の領域はヘッセ行列の固有値を利用した画像強調手法により管状構造を強調[14)]することで，後者は３次元回転型最小方向差分フィルタによる塊状領域を強調[15)]することで，それぞれ抽出される．このような表示法により，手術計画立案のための，血管・リンパ節の表示を行っている例を **Fig. 3** に示す．

３．２ 単臓器変形シミュレーション型

1) 概説

先述のように，このタイプのシミュレーションは，肝臓といった単一の臓器の変形のシミュレーションを行うものである．ここで重要となる技術は，(1) 臓器セグメンテーション，(2) 弾性モデル化，(3) 変形計算，(4) 可視化である．この分野に関しては数多くの研究が行われている．特に軟部組織の変形シミュレーションに関しては，高速有限要素法による計算 [16]，質点ばねモデルによる変形計算，球充填モデルによる変形計算など数多くの研究がある，代表的な研究としては，Cotin らによる軟部組織変形シミュレーション [8]，鈴木らによる球充填モデルによる肝臓変形・切開シミュレーションある [9]．以下，筆者らが開発を進めている仮想胃展開像生成システムについて述べる

2) 仮想胃展開像生成

この手法は，３次元 CT 像から取り出された胃領域情報を基に仮想的な胃展開像を生成するものである．日本の胃がん診断においては，胃展開標本が標準となっており，この展開標本上で切除範囲の評価を行っている．このような胃展開標本作成を３次元 CT 像に基づきシミュレーションできれば，最適な切除範囲の決定を仮想像上で行うことができ，手術計画立案に極めて有効である．

この仮想展開像標本生成手順法は以下の通りである [17]．まず胃壁領域を３次元 CT 像から抽出し，領域をなす各ボクセルの各頂点に質点を配置し，各辺と対角線にばねを張ることで弾性モデルを構築する．次に大湾側に沿って仮想的に切開線を入れ，切開線上に胃を平面状に展開するためのばねを張る．ばねに働く力を Newmark-β法などの逐次的計算法によって求め，これに基づき胃壁領域の変形を計算する．胃壁領域の変形結果に基づき原画像を変形し，それを可視化することで胃仮想展開像を得る．展開像生成シミュレーション結果の例を **Fig. 4** に示す．

３．３ 統合型手術治療シミュレーション型

1) 概説

このタイプのシミュレーションシステムは，内視鏡下手術そのものをシミュレーションするものである．先述のようにこのようなシミュレータとしては，(a) 患者毎モデル型と(b)一般(汎用)モデル型の２つがある．後者のシステムは力覚フィードバックを有した腹腔鏡下手術シミュレータが商品として販売されており，腹腔鏡下手術トレーニングなどで利用されている [11]．筆者らの研究グループでは，患者毎モデルに基づく腹腔鏡下手術シミュレータの実現を目指して，仮想腹腔鏡像生成シミュレーションシステムの開発を行っている [18]．以下のその概要を示す．

2) 仮想腹腔鏡像生成

この手法は，３次元 CT 像を基に仮想腹腔鏡像を生成する．腹腔鏡下手術は，気腹針により腹腔内に空気（二酸化炭素ガス）を注入し腹壁を持ち上げることで，手術に必要なスペースを作成し，直径 1cm 程度の穴を 3~5 個開け，腹腔鏡・鉗子等を挿入し，腹腔鏡から得られる画面を見ながら手術をするものである．この手術法は大きな開腹を伴わず患者にとって低侵襲であるが，手術に用いられる腹腔鏡の視野と移動範囲が限定されており，医師が術中に観察できる範囲は非常に狭い．そのため術中に重要な情報，例えば病変と腹部臓器との位置関係などを把握することが難しい場合があり，患者毎の腹腔鏡下手術シミュレーションシステムの開発が求められている．気腹がシミュレートされた仮想腹腔鏡像を作成することによって，術前に病変と周辺臓器の位置関係，内視鏡の挿入方法，腹腔鏡下手術に特有な鉗子などを検討可能となり，非常に有用であると考えられる．

このような腹腔鏡下手術を３次元 CT 像を利用してシミュレーションする場合，３次元 CT 像が気腹前の状態で撮影されていることが問題となる．仮想腹腔鏡像生成手法では，仮想的に気腹圧を発生させ腹壁を持ち上げ，また下に存在する臓器を変形させることで，腹腔内にスペースを作成し，腹腔鏡画像のシミュレーションを行う．具体的には，入力画像である３次元 CT 像から変形の対象となる腹壁領域を抽出する [19]．次に，抽出した腹壁領域からバネモデルを生成し，力学的な変形操作を行なうことで腹壁を持ち上げる．変形後のモデルと変形前のモデルとの対応関係を用いて CT 像を再構成する．最後に変形された CT 像を可視化することで仮想腹腔鏡像を得る．**Fig. 5** に仮想腹腔鏡像を生成した例を示す．

４．むすび

本稿では，内視鏡手術治療のシミュレーションについて，人体内部を内視鏡で観察可能な仮想化内視鏡から実際の腹腔鏡手術をシミュレーション可能な手法まで，医用画像処理技術の観点から概説した．臨床の場において，これらのシミュレーション手法がより一層利用されることが期待される．

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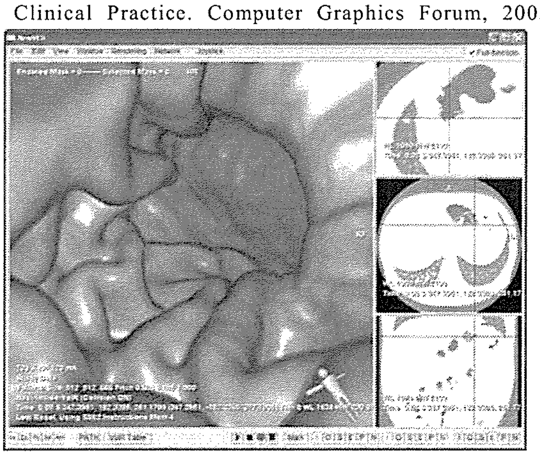

Fig. 1 Screenshot of virtual endoscopy system

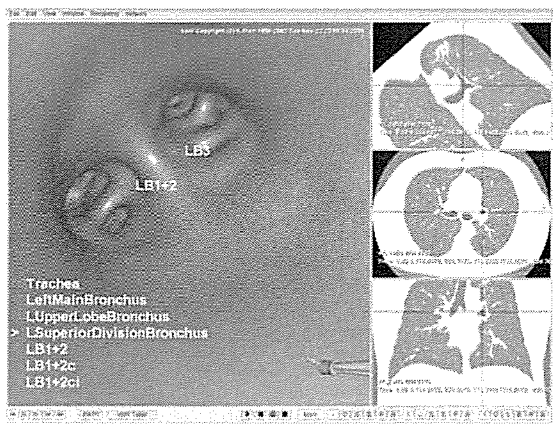

Fig. 2 TBLB simulation on virtual endoscopy system

Fig. 3 Blood vessels (red and green (selected vessels) lymph nodes (blue) dosplay for surgical planning on virtual endoscopy system (in collaboration with Dr. Nawano and Dr. Ito of NCC East).

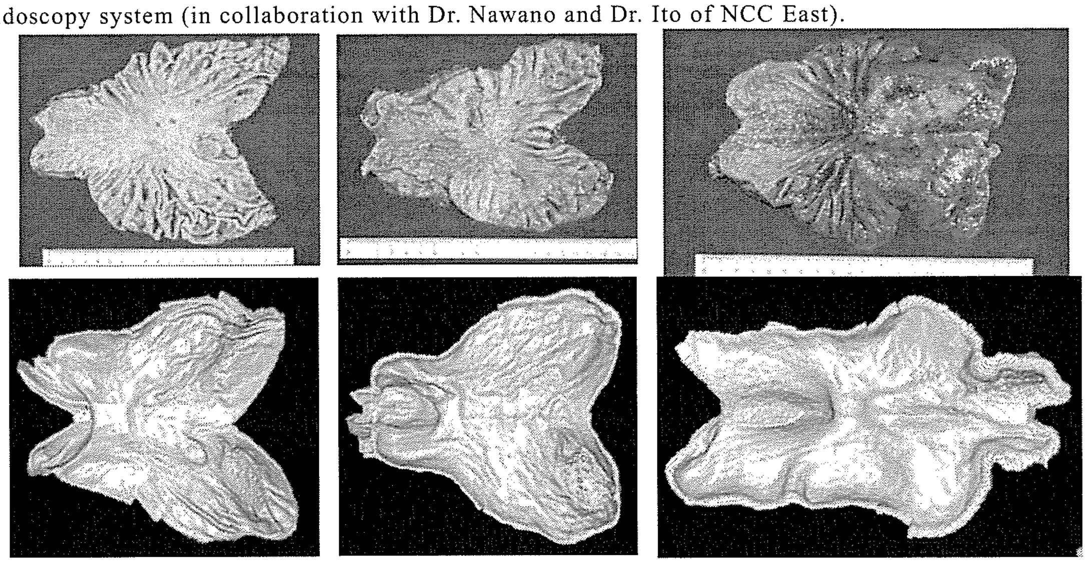

Fig. 4 Examples of virtual virtual unfolded view generation of the stomach. (upper row) real specimen. (lower row) virtual unfolded views.

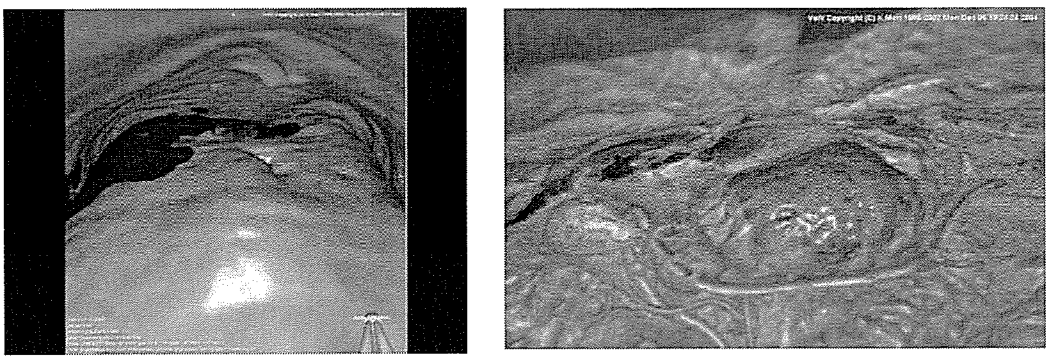

Fig. 5 Examples of virtual laparoscopic images (in collaboration with Dr. Hashizume of Kyusyu University and Drs. Fujiwara and Misawa of Nagoya University)

Transferring Bioelasticity Knowledge through Haptic Interaction

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This study establishes a practical environment for transferring knowledge on bioelasticity between expert and trainee medical practitioners. Through haptic interaction with a deformable virtual anatomical model, experts set the model's elasticity conditions by simulating a surgical procedure. Trainees experience the elasticity by attempting the same surgical manipulation.

Knowledge of soft-tissue bioelasticity is essential to medical practitioners in making a successful diagnosis and skillfully carrying out surgical procedures, such as palpation or handling surgical tools. Intraoperative palpation, for example, lets clinical medical practitioners determine the medical status of the body's organs, so it's essential that practitioners understand soft tissue's physical characteristics. Currently, general bioelasticity training is conducted in a "learn by doing" approach during daily medical work. This approach, the traditional style of clinical education, offers few chances for practice in clinical fields. Consequently, it's difficult for students to gain systematic experience in dealing with a wide variety of diseases or rare cases.

In this article, we describe how we constructed haptic media to represent the specific physical behavior of a beating aorta for instructing intraoperative palpation in cardiovascular surgery. We analyze the results of a communication sequence during a user study that involved skilled cardiovascular surgeons and students who trained on haptic interaction using our virtual, deformable models. The study tested this method's effectiveness in communicating both key elasticity properties and the means of manipulation required to master palpation. It's difficult to determine an aorta's bioelasticity, especially in cases such as sclerosis. Therefore, both the approach we created and the quantified knowledge gained during our study will be useful indices for the future development of haptically valid anatomical models.

We chose this area to investigate because currently, there's no training environment provided for intraoperative palpation despite its being a basic technique that all cardiovascular surgeons need to master. Another reason is that the aorta palpation procedure is simple and is performed with a simple push of a finger. Because current force-feedback devices, such as SensAble Technologies' Phantom (http://www.sensable.com), ably support the push operation, the only additional requirement for creating a realistic interaction environment is a valid physics-based aorta model.

Background: Surgical training methods

Virtual reality (VR)-based surgical simulators[1-7] have emerged as possible practical environments for residents to attempt repetitive training in surgical procedures. Physics-based models have been developed for simulating visual and haptic feedback of virtual organ manipulation.[8-15] Recent studies report that simulator-based training helps improve the results of actual surgery.[16,17]

Some of these haptic simulators, targeted specifically at palpation training, are designed to support bioelasticity knowledge acquisition.[4,5] In fact, we too have developed palpation simulators,[6,7] which concentrate on modeling beating behavior and organ–organ interaction in human bodies. In general, these simulators require more accurate haptic displays than the training simulators used in tool manipulation, such as suturing.[2,3,13]

More work, however, remains to be done to advance palpation simulators so they can be used effectively in training. The main obstacle lies in developing valid anatomical models because of the difficulty in accurately specifying organs' elasticity. Output from simulations varies accord-

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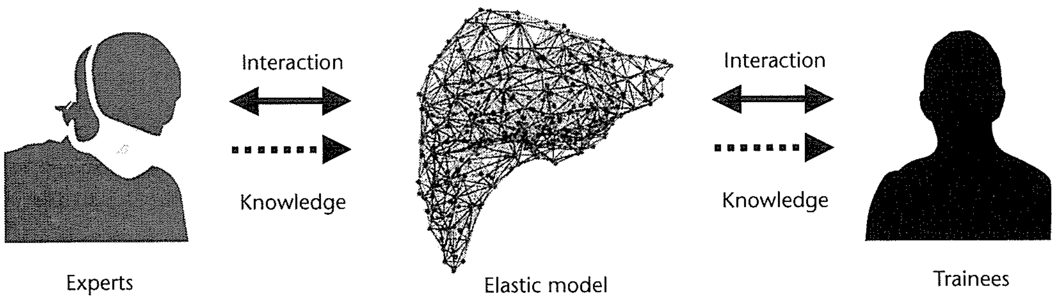

Figure 1. Basic concept. Elastic models support communication between experts and trainees as deformable media. Experts transfer their elasticity knowledge onto the models and trainees study the elastic behavior through haptic interaction.

ing to physical parameters and the complexity of the virtual organs that are being manipulated. Techniques for modeling patient-specific elasticity haven't yet been established; consequently, it's difficult to create accurate models for diseased organs. In addition, the effectiveness of training using haptic simulators must be further evaluated so that researchers can understand and improve simulator performance.

On the other hand, instruction by skilled medical practitioners has traditionally played a large role in the education of patient-specific elasticity. In practical situations, experts sometimes communicate images that are difficult for residents to conceptualize by referring to similar, known elastic objects. This approach is an accepted method of teaching, but verbal communication alone, without hands-on experience, isn't sufficient for effectively transferring knowledge on tissue elasticity. Therefore, an optimized education process requires a combined approach using both simulation and skilled instruction. If an expert's knowledge on bioelasticity could be efficiently transferred to trainees via a computer-assisted environment, trainees would learn key elasticity skills, including those needed for disease and rare-case scenarios.

Researchers have made some headway in developing haptic communication between experts and trainees. Haptic teaching systems developed in other fields focus mainly on tool manipulation[18,19] and are used to teach trajectory and applied force, especially in the field of handwriting. Satoshi Saga and colleagues[19] have proposed a haptic video system, which replays the force history acquired from an expert's manipulation. These approaches support the instruction of tool manipulation rather than provide a simulation environment where objects are manipulated. In manipulating elastic objects, such as during a surgical procedure, a surgeon feels reaction force other than that resulting from the tool's position. Consequently, a surgeon's education in physical behavior of elastic objects must include both haptic interaction and physical modeling of the object under manipulation.

In our study, we aimed to establish a communication support environment for transferring knowledge on bioelasticity using virtual anatomical models. Figure 1 shows the basic concept. We propose an interactive system that lets skilled medical staff instruct residents or medical students in organ bioelasticity by haptic interaction with the models. In our deformable media approach, virtual models are used to transfer elastic information from experts to trainees. The elastic information is stored quantitatively as part of the model's physical parameters. In a simple process, experts set up the model according to their experience, and trainees learn how to perceive the elastic information by attempting surgical procedures on the same model.

Supporting communication on bioelasticity

Medical practitioners generally learn objects' elasticity through attempting to recognize their physical behavior by conducting haptic interaction maneuvers, such as direct hand manipulation or varying the manner of contact. Practitioners also learn bioelasticity properties associated with specific diseases by touching and manipulating tumors in daily medical work.

Basic principles

To advance current training methods and techniques, practitioners require a platform on which medical procedures can be practiced. This platform—haptic media—should serve to let experts share their knowledge and trainees conduct self-study in the virtual space. Here, we outline the basic design, and the two primary tasks, of our proposed environment:

1. *Expert sets virtual model elasticity*. Experts simulate medical procedures on the virtual

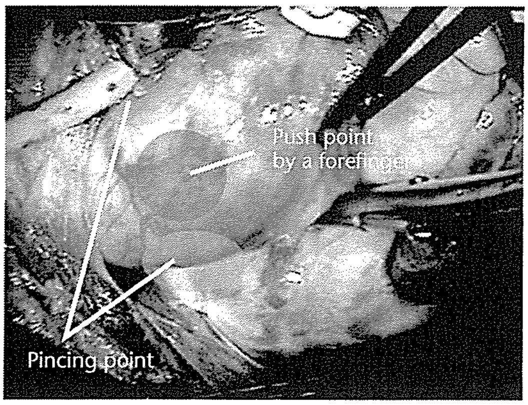

Figure 2. Intraoperative palpation in total aortic arch replacement surgery. Surgeons push or pinch the aorta's surface wth one or two fingers to identify the sclerotic region for diagnosis and overall strategy planning.

model. The feedback reaction force and deformation help the expert perceive the model's elasticity and compare with previous experience. If the elasticity differs from the expert's expectation, the expert changes the model's physical parameters. When the model's physical behavior matches the expert's expectation, the elastic parameters are stored, which completes this task.

2. *Trainee learns virtual model elasticity*. In this task, trainees experience the model's elasticity by repeatedly performing the same manipulation that the experts carried out in the first task. Over time—which might involve a trainee's conducting the same procedure once every few days—the trainee acquires a stable perception of different conditions, and recognizes the elasticity relating to a specific situation.

Virtual aorta palpation system design

For discussion purposes, we've considered a total aortic arch-replacement surgery. In such a surgery, surgeons must perform intraoperative palpation of the aorta to establish the tissues' elasticity. Surgeons do so because they need to identify the sclerotic region to determine the overall surgical strategy required to deal with the tumors (see Figure 2). Because the nature of the sclerotic region cannot be ascertained from visual information, such as texture and 3D shape alone, surgeons—using one or two fingers—push or pinch several points of the aorta.

To provide practitioners with an interactive environment enabling them to rehearse realistic palpation procedures, we first need a clear description of how interaction with the virtual model should occur. Furthermore, it's essential to engineer realistic visual and haptic feedback. Specifically, the aorta model must aptly simulate both autonomous beating and volumetric distribution of elasticity equivalent to that of the sclerotic status of real-life tissue. Additionally, to accurately model the surgeon's manner of palpation, the haptic interface should support both one- and two-finger manipulation. Finally, the interactive modeling interface should allow surgeons to modify physical parameters based on their empirical bioelasticity knowledge.

On the basis of these requirements and additional discussions with cardiovascular surgeons, we've designed a virtual aorta palpation system (see Figure 3). The system visualizes a 3D aortic arch model as a polygonal object in the virtual space. Users control two sphere-shaped manipulators via two Phantom haptic devices—model Premium 1.5. The manipulators represent the 3D positions of a user's thumb and index finger, respectively. The Phantom's collision detection algorithm responds to any interaction between the manipulators and the model, while the physics-based simulation algorithm calculates the reaction force and deformation using the contact points and manipulators' current positions. The Phantom device conveys the calculated reaction force to the user and displays the deformation result on the screen as a transformation of the 3D model. The Phantom haptic device also lets surgeons interactively update the model's elasticity and boundary conditions when the physical behavior differs from the surgeons' expectations. We provide a graphical user interface that supports elastic modeling and manages given physical parameters.

To simulate the physical behavior of a beating aorta while maintaining a haptic-compatible refresh rate, we propose new finite-element-based computation methods. Although many studies have concentrated on developing various kinds of physics-based models,[8-15] the finite element method (FEM) is known to be the most accurate computational model for simulating the biomechanical behavior of elastic soft tissues. Although FEM-based simulation provides accurate and stable deformation, however, it carries a high calculation cost. To allow real-time interaction with a volumetric-deformable object, Morten Bro-Nielsen and Stephane Cotin have proposed a condensation technique.[9] This technique reduces the size of a stiffness matrix in the preprocessing stage and performs real-time simulation for detailed objects. More recently, Koichi Hirota and Toyohisa Kaneko have achieved fast computation of reaction force by using an efficient translation of a matrix calculation.[14]

Although most previous studies didn't try to simulate dynamic behavior like pulsation or beating status, our previous study detailed the 3D anatomical shapes and approximate time series pressure required to realistically represent the haptic feedback of heartbeats.[6] In the study, we've used these features to create haptic-deformable media to represent a beating aorta's dynamic behavior during cardiovascular surgery. Our proposed FEM-based computation methods then calculate the reaction force and biomechanical deformation that reflects the internal pressure induced by the user's manipulation.

Physics-based modeling of a beating aorta

Physics-based simulation requires a 3D shape with elastic information of the target organ.

3D shape and elasticity modeling

To construct our virtual aorta model, we acquired images of patients' aortas from computerized tomography or magnetic resonance imaging techniques. We extracted a 3D region of the aorta from voxels and divided them into finite tetrahedra. Each tetrahedron represented a part of the 3D region of the aorta and also its particular physical parameters: Young's modulus and Poisson's ratio, which are measures of elasticity. We used the volumetric grid topology and physical parameters to calculate a stiffness matrix, enabling finite-element-based simulation. We categorized all vertices, based on three different boundary conditions, into fixed vertices, internal wall vertices, and other free vertices. Fixed vertices represented tissue that connects or contacts other organs. Internal wall vertices represented the aorta's beating, holding dynamic force by a time series of blood pressure data. We represented the condition of arterial sclerosis by setting a high Young's modulus to the desired region.

We created a virtual 3D shape of a normal aortic arch from the Visible Human Male data set.[20] Using Mercury Computer Systems' Amira 3.1 modeling software (http://www.amiravis.com), we reconstructed the aorta surface and generated tetrahedral grids. Figure 4 illustrates the constructed aortic arch model. The total number of the vertices was 1,651. The edge vertices of the 3D model, which simulated connection to the heart and other vessels, were defined as fixed because these areas didn't move in actual palpation. Each vertex was represented as a small sphere colored according to its boundary condition, shown in Figure 4.

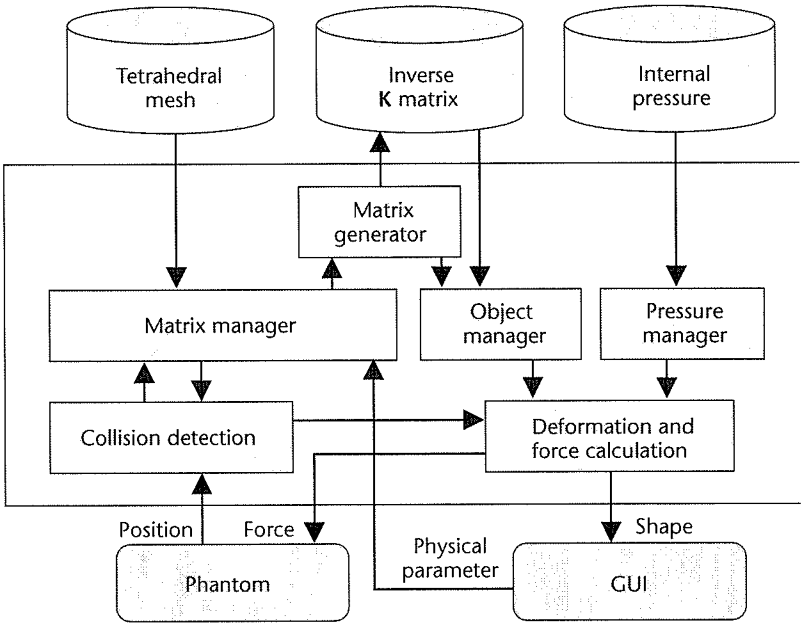

Figure 3. Overall framework for the simulation system developed for instructing palpation of the aorta. Surgeons and residents used this system to rehearse realistic palpation procedures on the virtual aorta model. Skilled surgeons conveyed their bioelasticity knowledge by modifying the model elasticity.

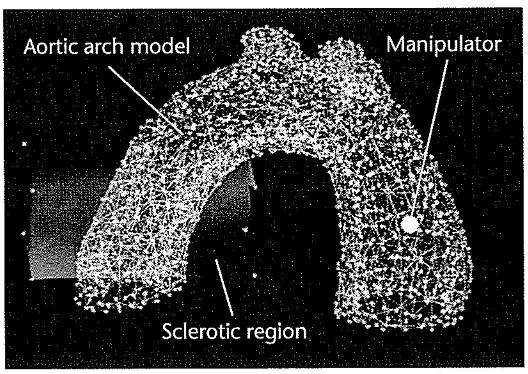

Figure 4. The aortic arch model. The 3D shape was acquired from the Visible Human Male data set. A high Young's modulus was given to tetrahedral elements in the sclerotic region using a 3D bounding box. The colors of vertices represented boundary conditions. Red vertices were "fixed," green vertices were "internal wall," and blue vertices were "free" for deformation.

The system we developed also provided an elastic modeling interface to support flexible parameter setting by skilled surgeons. The view of the virtual space contained a 3D bounding box (see Figure 4) whose shape and 3D position were controlled interactively using a slider bar and edit box in the graphical user interface. We also prepared several shape templates such as spheres and cubes. This interface let surgeons modify the model elasticity and boundary conditions as needed. Following this interactive elasticity modeling, the system automatically updated the stiffness matrix by running matrix generation algorithms.

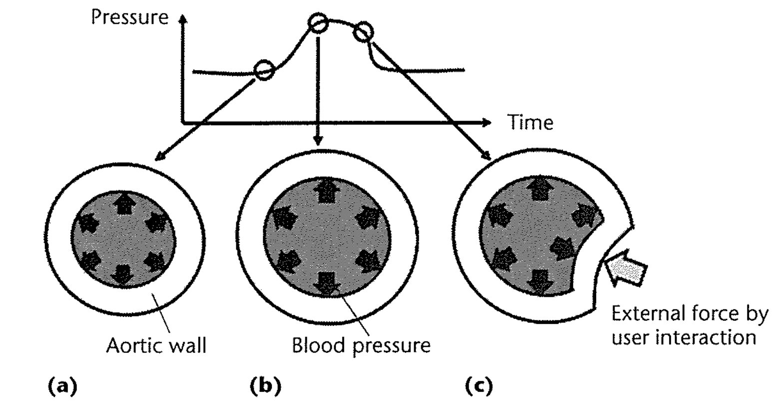

Figure 5. FEM-based computation model for a beating aorta: (a) 3D shape; (b) aorta expansion after higher pressure exerted; and (c) the result of external force applied.

Haptic interaction

The collision-detection algorithm, in conjunction with physics-based simulation, enabled virtual palpation to be performed on the aorta model. The algorithm handled the intersection between the model's fingertip manipulators and the tetrahedral grids, for which we employed proxy-based haptic rendering methods.[21,22] The algorithm was based on point-polygon collision detection, which, despite being a simplification of a real aorta, was sufficient to handle the basic interaction of a surgeon's fingertips touching the aortic wall in real surgery. The main challenge of this study was to simulate and describe interaction with a deformable object that's subject to autonomous beating.

Figure 5 outlines the basis of our haptic interaction model for aorta palpation. A 2D cylinder cross-section approximately illustrates the aortic wall. When blood pressure was applied to the wall (depicted in Figure 5a), the FEM-based simulation responded by expanding the 3D shape. The higher the pressure applied, the larger the expansion, as Figure 5b shows. The system provided simulation results at discrete time steps, effectively modeling the aorta wall's autonomous beating. In addition, when an external force initiated a small displacement in the aorta model, the reaction force was calculated and conveyed to the user's fingertips via the Phantom devices, indicated in Figure 5c.

Cardiovascular surgeons reported that, during a palpation procedure, they avoid pinching the aortic wall too hard to avoid damaging soft tissues and to minimize the aorta's deformation. Therefore, our proposed model assumed that internal blood pressure wasn't affected by surgical palpation, evidence that linear finite element models were a good approximation for real-life aorta palpation.

FEM-based computation

We based our calculation method—to simulate the reaction force and deformation of a beating aorta—on linear elastic theory. Assuming that the internal force is in equilibrium at each discrete time step, the relationship between external force and displacement on an elastic object is defined by $f = \mathbf{K}u$ where, on all vertices, u is displacement and f is external force, including blood pressure. K denotes the stiffness matrix of the object constructed by grid topology and physical parameters (Young's modulus and Poisson ratio). The K matrix can be efficiently reduced by condensation[9] and elimination of the fixed vertices in the preprocessing stage. This expression is simplified to $u = \mathbf{L}f$ using the condensed inverse stiffness matrix L, which defines the physical relationship between the external force f and the displacement u on the surface vertices.

To represent autonomous beating and to simulate the physical behavior of the user's interaction, we divided the surface vertices into three groups: contacted vertices, internal wall vertices, and other free vertices. Contacted vertices are directly displaced by the user's manipulation. Internal wall vertices are affected by the time series of blood pressure. Equation 1 expands $u = \mathbf{L}f$ by using the initial letters of categorized vertices to represent the coefficients of the matrices.

$$\begin{pmatrix} u_i \\ u_o \\ u_c \end{pmatrix} = \begin{pmatrix} \mathbf{L}_{ii} & \mathbf{L}_{io} & \mathbf{L}_{ic} \\ \mathbf{L}_{oi} & \mathbf{L}_{oo} & \mathbf{L}_{oc} \\ \mathbf{L}_{ci} & \mathbf{L}_{co} & \mathbf{L}_{cc} \end{pmatrix} \begin{pmatrix} f_i \\ f_o \\ f_c \end{pmatrix} \tag{1}$$

where f_i denotes blood pressure that's applied to the internal wall vertices and u_c is displacement of the contacted vertex manipulated through the Phantom haptic device. Considering that f_o is constant zero, the relationship between u_c, f_i, and f_c is described as $u_c = \mathbf{L}_{ci}f_i + \mathbf{L}_{cc}f_c$. Consequently, f_c is given as $f_c = \mathbf{L}_{cc}^{-1}(u_c - \mathbf{L}_{ci}f_i)$.

Accordingly, f_c is external force on the contacted vertex, and $-f_c$ is reaction force conveyed to the user. Note that we can obtain $\mathbf{L}_{cc}$ and $\mathbf{L}_{ci}$ by precomputation because both are defined by Young's modulus and Poisson's ratio. We calculate $\mathbf{L}_{cc}$ and $\mathbf{L}_{ci}$ for all free vertices and perform a refresh rate of more than 1,000 Hz to maintain stable force feedback. The dynamic transition of f_c at discrete time steps lets us present haptic feedback of the beating aorta under the effects of time series blood pressure. Applying f_c to Equation 1 provides the displacement u_o on other free vertices.

Evaluation and user study

In this section, we describe the virtual palpation system's operation, example results, and outputs. We then present the design of studies that evaluate the effectiveness of our system in supporting communication on bioelasticity.

System verification

We processed the computation algorithms on a standard PC with a Pentium 4 2.4-GHz CPU and 1,024 Mbytes of memory. We optimized the matrix calculation on the CPU with Intel's Math Kernel Library.

Our first step was to confirm the visual and haptic quality and performance of the developed system with cardiovascular surgeons. In this experiment, we set Young's modulus of the normal model to 1.0 megapascal (MPa), that of the sclerotic model to 3.0 MPa, and the Poisson's ratio to 0.48, based on the measured data for a normal aortic wall that we'd obtained previously.[23] Later, we explain how we tested the model's physical characteristics, compared to the empirical knowledge of skilled cardiovascular surgeons.

Figure 6 shows how the user interacted with the virtual system and gives an example of the deformation that occurred in the virtual aorta following pinching with two fingers. The user touched two manipulators in real space with the tips of the Phantom devices that were worn on the fingertips (bottom-left image). These manipulators translated the position of the fingertips to the model aorta in the virtual space. In Figure 6, we highlight the virtual position of the fingertips (top-left image) that corresponded to the fingertips' real position (bottom-left image).

Next, we tested the accuracy of the model in simulating both normal and sclerotic aorta conditions. The proposed FEM-based model simulated reaction force and deformation for an aorta subjected to time series pressure. Figure 7 shows the relationship between a dynamic transition of the reaction force and the applied time series blood pressure when the aorta wall was subjected to a specific displacement by the fingers. We programmed two regions of the virtual aorta, each with a different state (normal and sclerotic). The two different regions showed different absolute values of reaction force and beating status for the same given displacement. The sclerotic part did not reproduce the pulse with as great a magnitude as the normal part. These phenomena were similar to that observed by Haruo Okino and colleagues.[23] Thus, our model simulated realistic haptic feedback reflecting the dynamic physical behavior of both a normal and a sclerotic aorta wall.

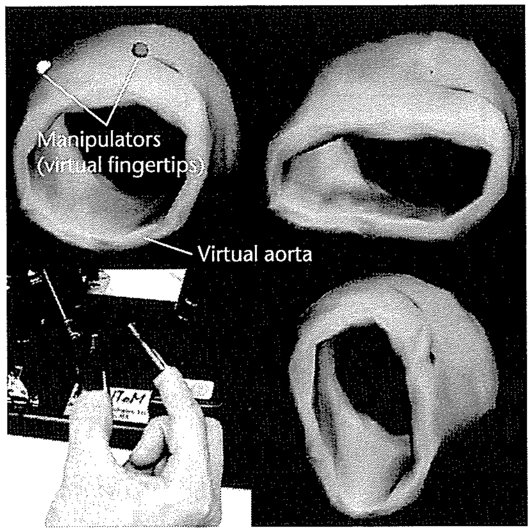

Figure 6. Real-time deformation in palpating the virtual aorta. Two manipulators controlled by two SensAble Technologies' Phantom haptic devices enable users to pinch the model interactively.

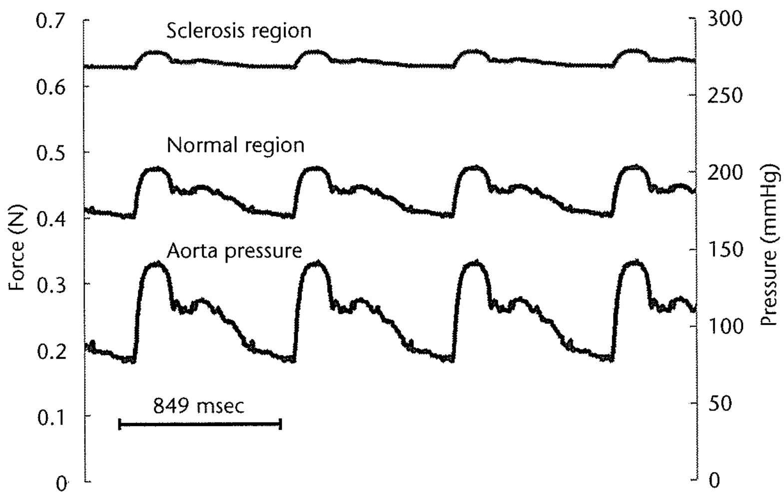

Figure 7. Reaction force in pushing normal/sclerotic region on a virtual aorta sclerotic model. This simulation result demonstrated that the sclerotic part did not reproduce the pulse with as great a magnitude as normal tissue.

The developed model had 1,651 vertices, which provided sufficient visual quality to represent the 3D shape of an aortic arch. In this case, the calculation time was 0.08 ms for force feedback and 0.5 ms for deformation. These times confirmed that the proposed calculation methods achieved a sufficient refresh rate to handle deformable virtual media with autonomous beating.

User tests

As we explained, the system's role was to support the communication of knowledge on aorta bioelasticity. We categorized this communication in three separate procedures and planned three user studies to test how well the system supported this communication, basing each experiment on each of the following hypotheses:

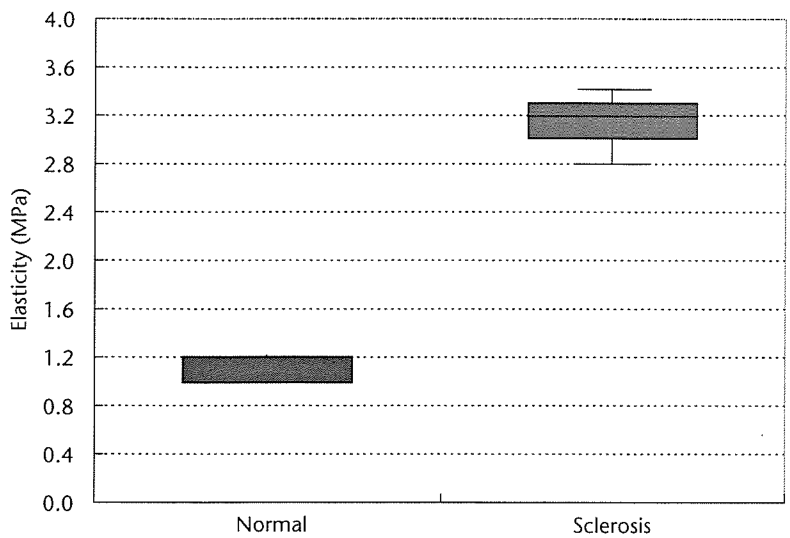

Figure 8. Interactive modeling results of normal elasticity and the threshold elasticity that should be regarded as sclerotic. The box plot indicates 25 percent (minimum) and 75 percent (maximum) of the data. This demonstrates that skilled surgeons are able to recognize key aorta palpation bioelasticity and also show their knowledge on bioelasticity in aorta palpation.

- *Elasticity modeling.* If experts knew the aorta's correct tissue elasticity during palpation and the system provided a valid elasticity-modeling environment, we could set up the virtual aorta with identical given elasticity.
- *Elasticity recognition.* If haptic interaction with the virtual media effectively supported communication on bioelasticity, trainees would be able to recognize the given elasticity more easily than with other training methods.
- *Learning elasticity.* If the system supported instruction of a specific bioelasticity, trainees would be able to learn the given elasticity through continuous repetition of the palpation procedure.

We focused on two elasticity conditions that were considered key in intraoperative aorta palpation: normal tissue elasticity and the threshold elasticity that's regarded as sclerotic. Cardiovascular surgeons have to learn both elasticity conditions to identify the sclerotic region and determine the overall surgical strategy. In the next three sections, we describe the results of the user study.

Elasticity modeling

First, we examined the capability of the system to carry out elasticity modeling of the virtual aorta, working with eight skilled surgeons from the Department of Cardiovascular Surgery at Kyoto University Hospital. We prepared 20 aorta models with the same 3D shape (shown in Figure 4) but different levels of stiffness. We set the uniform elasticity of each model by changing the input parameter of Young's modulus, which varied from 0.2 MPa to 4.0 MPa. The Poisson's ratio was set as 0.48. In advance of the experiment, we allowed the surgeons a few minutes of practice time during which they attempted the palpation procedure at leisure to accustom themselves to the virtual environment.

For the study itself, the surgeons conducted a normal palpation procedure on the 3D virtual aorta. If the physical behavior and reaction force differed from their expectation of how a normal aorta should have behaved, we changed the model elasticity accordingly. They repeatedly palpated the models and selected the one that most realistically simulated the elasticity of a normal aorta. The same procedure was used to determine the threshold elasticity that should have indicated sclerosis in real surgery. We randomized the order in which the surgeons tested the different models. The same palpation point on the model was used for all surgeons.

Figure 8 shows the statistical results for the models that were selected for normal and sclerotic elasticity conditions. The left-hand graph shows there were only two different models, 1.0 MPa and 1.2 MPa, which the surgeons deemed to correspond to their experience of a normal aorta. Specifically, five surgeons selected the 1.0-MPa model; three chose the 1.2-MPa model. Next, the right-hand graph in Figure 8 shows the results of selection for the sclerotic aorta. The graph suggests that anything over 3.0 MPa corresponded to sclerosis, according to the experience of the surgeons. These results lead us to the following insights:

- A virtual aorta with a Young's modulus between 1.0 MPa and 1.2 MPa effectively displays the physical behavior of a normal aorta.
- A virtual aorta having a Young's modulus of over 3.0 MPa simulates sclerotic status.
- All skilled surgeons recognize the absolute elasticity of normal and sclerotic conditions by touch. This means that elasticity is the most important information to communicate when trainees try to master the palpation of an aorta.

The surgeons who evaluated our system reported that manipulating the virtual model using the two Phantom devices was comparable to the experience of real surgery. Also, they stated that they experienced a realistic reaction force following adjustment of the model's physical parameters. In addition, a sufficiently small disagreement arose among the surgeons in choosing the models that best represented normal and sclerotic conditions of the aorta. These results indicate that our system enabled virtual palpation that effectively mirrored the real-life procedure.

Elasticity recognition

As shown in the first experiment, the ability to recognize the elasticity of normal and sclerotic aortas is important. In our second experiment, we aimed to confirm that haptic interaction with a deformable media was useful for bioelasticity training. We initiated a study with 18 medical students who hadn't experienced palpation of a real aorta and examined their ability to recognize different elasticity conditions.

Ten of the virtual aorta models used in the previous experiment were again prepared and the students performed virtual palpation in the same manner. Again the elasticity ranged from 0.2 MPa to 2.0 MPa. Some time was given for the students to become accustomed to manipulation and the manner of touching the model in virtual space. Then, each student was asked to select the model, which they thought best represented a real-life normal aorta. We prepared the students for making their selection under the following conditions.

Condition 1: No information. The medical students had only experienced the elasticity of a cadaver's aorta during their medical training and had not received any specific instruction on aorta elasticity prior to starting the experiment. Because the elasticity of a cadaver is totally different from that of a living body, they had no experience of an in vivo aorta and, in selecting from among our models, had to imagine what a real aorta would feel like.

Condition 2: Verbal information. The students underwent instruction from cardiovascular surgeons who described the physical characteristics of a normal aorta using both brief explanation and rubber hoses as a physical representation of aorta elasticity. This technique is a conventional teaching method.

Condition 3: Haptic instruction. The medical students were given 1 minute to become familiar with the elasticity of a normal aorta by carrying out virtual palpation on our developed system. The 1.0 MPa value was specified as the Young's modulus of this "normal" virtual aorta.

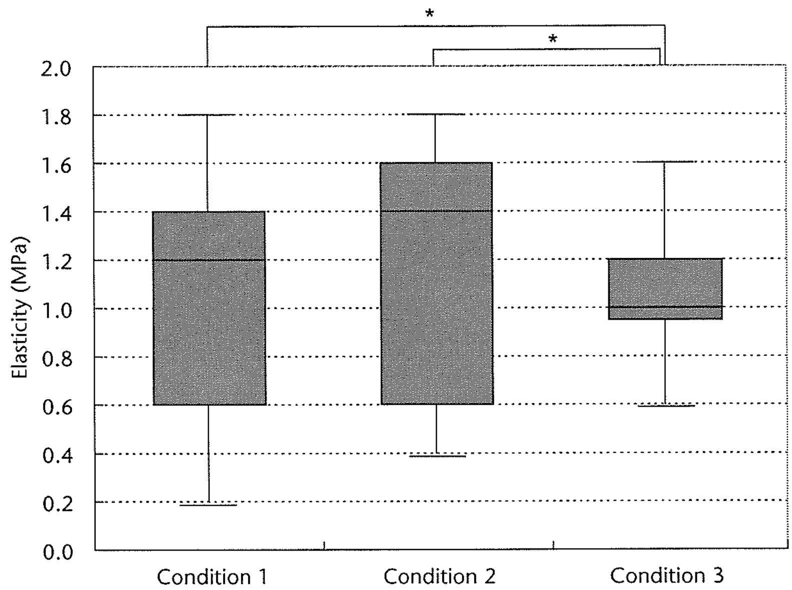

Figure 9. Statistical results of elasticity recognition of a 1.0-MPa normal aorta model in three cases: conventional instruction (condition 1 or 2) and simulator-based instruction (condition 3). (The asterisks indicate significant statistical difference between the two data values.)

Figure 9 illustrates the statistical results for the elasticity of the models that the students selected. The statistical analysis (F-test) result shows significant difference between the conventional means of instruction (condition 1 and condition 2) and simulator-based instruction (condition 3). The distribution of the graph at condition 2 still shows a large spread. This result shows that verbal communication alone isn't an effective means for students to learn the elasticity of a specific aorta condition, because students have an existing expectation of the elasticity that can't be altered simply by hearing what it should feel like. The graph of condition 3 is close to a normal distribution, and more than 80 percent of the students selected models with elasticity between 0.8 MPa and 1.2 MPa. This result demonstrates that haptic interaction using our system is an effective method for enabling students to recognize the normal stiffness of the virtual aorta model. Furthermore, the Young's modulus of 1.0 MPa had been configured by expert surgeons in the previous experiment, meaning that the students effectively experienced the elastic characteristics of an in vivo normal aorta through virtual palpation. We only conducted the second experiment employing the normal aorta because verbal representation isn't generally used to describe a sclerotic aorta in clinical work.

Learning elasticity

In this third, final experiment, we tested the effectiveness of learning elasticity through virtu-

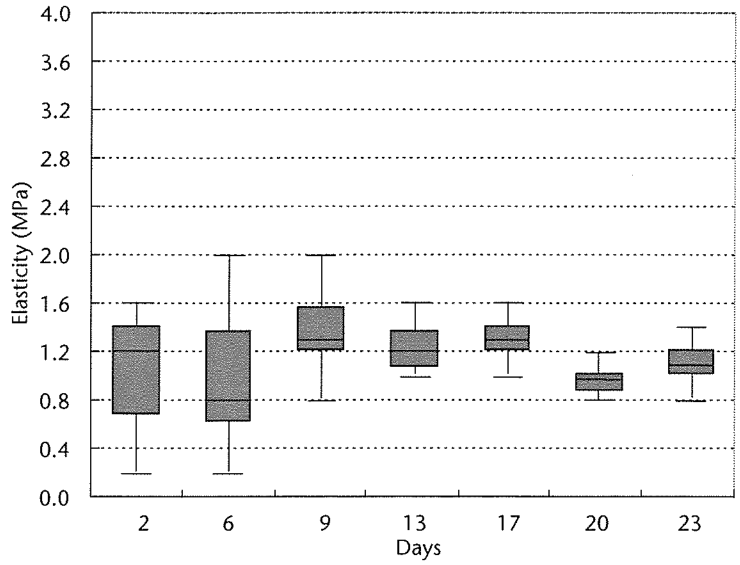

Figure 10. Virtual palpation learning curve for the normal model. The target was 1.0 MPa. The spread narrowed and the median selection approached 1.0 MPa over time.

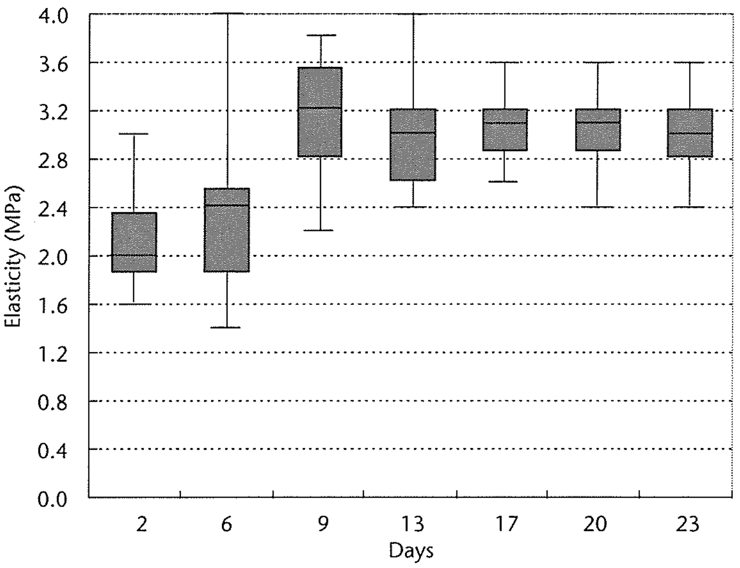

Figure 11. Virtual palpation learning curve for the sclerotic model. The target was 3.0 MPa. The spread narrowed and the median approached the target over time. Compared to Figure 10, each trial features a greater spread of selections.

al palpation. The second experiment had demonstrated that haptic instruction is an efficient means of enabling elasticity recognition, but to master palpation, students must memorize the critical elasticity they encounter during the procedure. Therefore, our study needed to further examine whether students were able to retain their knowledge of elasticity. In this last experiment, we aimed to provide students with a learning curve, in the form of a month-long course of training. We carried out the simulator-based learning based on the following conditions:

- The participants were 10 students who hadn't previously experienced a real aorta through touch.
- For each participant, seven trials were conducted, occurring approximately twice a week. The total experimental period was 23 days.
- Each trial consisted of two steps: testing and learning. Participants completed both steps at every trial.
- Test stage: Participants were asked to select the aorta models that they believed to be normal or sclerotic, from 20 aorta models with Young's modulus between 0.2 MPa and 4.0 MPa, based on their current bioelasticity knowledge.
- Learning stage: The correct normal (1.0 MPa) and sclerotic (3.0 MPa) aorta models were revealed. Examinees were given a few minutes to try to learn the elasticity of the models through virtual palpation.
- The palpation point on the aorta model was fixed throughout the experiment.

Figure 10 shows the selection data over the course of the third experiment in which the student participants attempted to choose the normal aorta. The spread of the participants' selections clearly narrowed over time, and the median eventually approached 1.0 MPa. Our statistical analysis (T-test) reveals a significant difference in the spread between the first and third trials, and the second and third trials. These graphs demonstrate that several separate training opportunities effectively contribute to learning the stiffness characteristics of the normal aorta.

Figure 11, which shows the selection data for the sclerotic condition, suggests that haptic instruction using the developed system is effective. However, there is a wider spread in the selections, compared to the learning curve in Figure 10. This tendency is consistent with what is known about human perception, in that human haptic sensitivity to relative physical behavior is proportional to the logarithm of absolute stiffness.

Final remarks

In evaluating the results of our experiments, we discuss how the efficacy of realistic VR-based palpation might be further improved. First, to achieve a real-time refresh rate, we simplified the physical behavior of an in vivo aorta by ignoring some essential functions. We didn't consider the effects of local pressure or blood flow, and simulated continuous autonomous beating with discrete time-series blood-pressure data. For noncomplex 3D aorta models, the blood pressure applied to the internal walls was assumed to be constant and independent of local position. The advantage of this simplification was that it aided fast calculation for valid haptic feedback and let us reproduce key soft-tissue behavior (for example, pulsation and elasticity) in virtual palpation. Our model achieved a refresh rate of more than 1,000 Hz in the reaction force calculation owing to our simplification of boundary conditions and inverse matrix calculation.

Another simplification was to reproduce the dynamic force of autonomous beating using a static FEM model. The variance in elasticity between the model and real life was small, and the surgeons who evaluated our system agreed that it demonstrated realistic force feedback and graphical deformation for the palpation procedure.

Although VR-based surgical simulators provide an effective training environment in clinical training, the representation and validation of specific bioelasticity behavior remains a problem. Both our proposed bioelasticity communication support environment and the quantified knowledge gained during our study will be useful indices for the future development of haptic anatomical models.

The approach we followed is potentially applicable to other virtual organ models that require a multiphysics simulation. We suggest that more detailed modeling, especially focused on a number of specific diseases and morbid stages, will be carried out as future work. This work will contribute to the compilation of practical instruction courseware. **MM**

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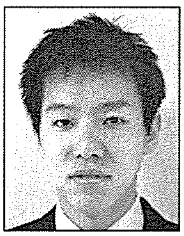

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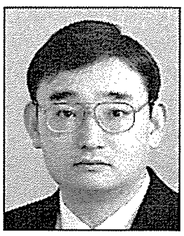

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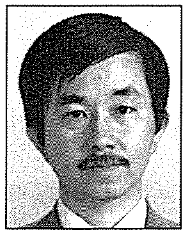

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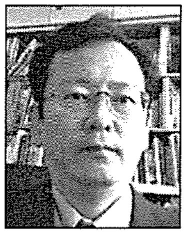

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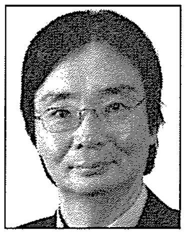

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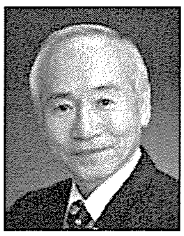

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Physics-based Manipulation of Volumetric Images for Preoperative Surgical Simulation

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ABSTRACT

Volume visualization software allows medical stuffs to extract meaningful information in patient specific surgical planning and diagnosis. In recent days, in order to discuss surgical strategy and to share the plan, their interest is not only visualizing 3D anatomical structure but simulating surgical procedures directly on the volumetrically rendered image. This study aims to establish interactive volume surgical simulation that allows surgeons to rehearse surgical procedure on patient's volumetric CT/MRI images. Physical phenomena between surgical instruments and organs like tumor resection and pinching/pushing manipulation are simulated with a FEM-based deformable model. We also consider interactive performance on standard PCs. This presentation reports details of the methods, the prototype surgical simulation system and some experimental results simulating surgical procedure in minimally invasive surgery.

KEY WORDS: Volume Manipulation, User Interface, Physics-Based Deformation, Surgical Simulation

INTRODUCTION

Volumetrically rendered images (called volumetric images in this paper) of patient 3D CT/MRI dataset are widely used in diagnosis and preoperative surgical planning. While the volumetric images are efficient to grasp 3D shape of organs and positional relationship of tumors, most current viewers can not simulate surgical process or surgical procedure like cutting, resection and pinching manipulation etc.

So far, some studies proposed basic techniques for volume edit [1-3] and physics-based deformation [4-5] on volumetric images. These methods are effective to represent volume cut or deformation interactively. However, sophisticated and practical simulation models are necessary to reproduce physical phenomena like pinching and ablation between surgical instruments and organs (see Fig. 1). Also, in order to support surgical planning by surgeons themselves, it is essential to provide user-friendly and intuitive interaction environment where surgeons can rehearse surgical procedure regarding volumetric images as virtual organs. Considering these features, development of practical interface and volume representation is required for enabling intuitive interaction with volumetric images on the simulation system.

This study aims to establish an advanced preoperative rehearsal system where surgeons try surgical procedure on patient's volumetric CT/MRI images. In this paper, we propose a volume manipulation framework based on real-time volume visualization and physics-based simulation technique. Unlike foregoing studies, the framework enables 6DOF surface constraint based manipulation of volumetric organ images. As the manipulator operated by a user is not a point but has 3D shape, realistic physical phenomena like pinching/grasping manipulation can be simulated. We also consider the system achieves interactive performance (over 30Hz refresh rate) on standard PCs, which is essential for interactively rehearsing surgical procedure. This paper describes details of the proposed methods, the prototype volume manipulation system and reports some experimental results.

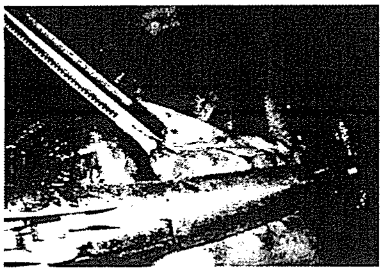

Fig. 1 Pinching manipulation with forceps in thoracoscopic surgery

PHYSICS-BASED MANIPULATION FOR PREOPERATIVE REHEARSAL

The developed framework supports user's interactive manipulation on volumetrically rendered images reconstructed from patient CT/MRI dataset. We describe such manipulation consistently as volume interaction between manipulators and organ objects (see Fig. 2). The geometry has a grid topology (e.g. tetrahedral mesh) generally used in finite element (FE) analysis and handles one elastic /rigid object like an organ, tissue and surgical instrument. The geometry of organs, for example, is created through general modeling process: volume segmentation, surface definition and grid generation.

Our framework deals with physics-based manipulation of tetrahedral grids while representing interaction results as volume deformation in real time. Note that updating large number of voxels is time consuming process and becomes serious drawback to interactive performance. The proposed algorithms do not update any voxels but visualize physical phenomena on the geometry by interpolating internal voxels using 3D texture mapping techniques.

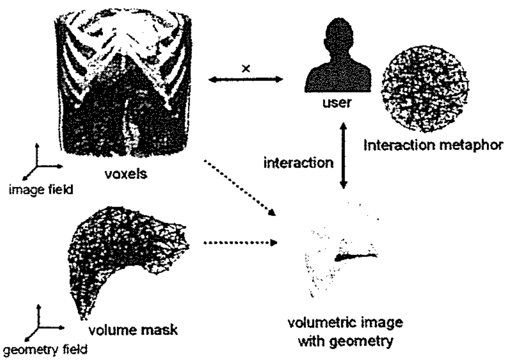

Fig. 2 Basic concept of our volume interaction framework

SURFACE CONSTRAINT AND DEFORMATION

In order to simulate surface constraints by surgical instruments, we developed a volume manipulation model which enhances applicability compared to conventional point-object interaction models. In this case, it is valid to assume organs are soft object and surgical tools are rigid. Fig. 3 briefly illustrates this interaction model and a deformation example. When intersection between the organ object and the manipulator is detected, the intersected area is regarded as grasped and displacement is applied to the vertices of the organ model.

More specific description is needed for the interaction between the organ object and the manipulator. Because we focus on surgical manipulation with forceps, the manipulator was currently modeled as simple two surfaces facing each other and a center point like in Figure 4. The virtual forceps is controlled through a 6DOF input device. The displacement *d* for each displaced vertex is obtained by the following equation.

$$\vec{d} = \vec{n} \cdot (q - p_0) \times \vec{n} \tag{1}$$

where n is the normal vector of the surface of the virtual forceps, q is the initial position of the displaced vertex and p_0 is a vertex on the surface.

If the position of the vertices is geometrically transformed while satisfying displacement by the virtual forceps surfaces, additional manipulation like both pinching and rotation is performed.

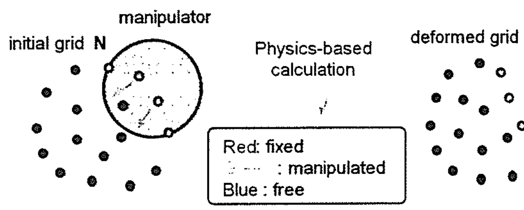

Fig. 3 Surface constraint based manipulation

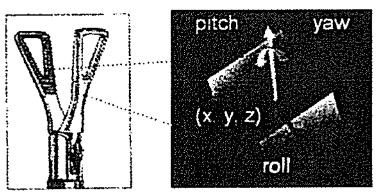

Fig. 4 A real forceps illustration and a basic virtual forceps model

Using the given displacement as boundary condition, we calculate physics-based deformation based on linear finite element formulation. Although non-linear representation is valid for accurate simulation, we currently focus on interactive performance for smooth animation.

$$f = Ku\,,\;\; L = K^{-1} \tag{2}$$

We divided the vertices into two groups: contacted vertices and other free vertices. Contacted vertices are directly displaced by the user's manipulation. Equation (2) expands $u = Lf$ by using the initial letters of categorized vertices to represent the coefficients of the matrices.

$$\begin{pmatrix} u_o \\ u_c \end{pmatrix} = \begin{pmatrix} L_{oo} & L_{oc} \\ L_{co} & L_{cc} \end{pmatrix} \begin{pmatrix} f_o \\ f_c \end{pmatrix} \tag{3}$$

where u_c is displacement of the contacted vertex manipulated through the haptic device. Considering that f_o is constant zero, the relationship between u_c, and f_c is described as

$$f_c = L_{cc}^{-1} u_c \tag{4}$$

f_c is external pressure on the contacted vertex. Applying f_c to Equation (3) provides the displacement u_o on other free vertices.

$$u_o = L_{oc} f_c = L_{oc} L_{cc}{}^{-1} u_c \tag{5}$$

Note that we can obtain the inverse stiffness matrix L by pre-computation because it is specifically defined by Young's modulus and Poisson's ratio of the elastic object. We re-calculate L_{cc} and L_{oc} for all free vertices when manipulated vertices are updated. Although the order of the calculation cost is $O(m^3)$ (m: the number of manipulated vertices), m is not large in representing pinching area. Moreover, since this approach does not need inverse matrix calculation during manipulation, real-time computation is possible.

VOLUME VISUALIZATION

Slice-based volume rendering of tetrahedral grid and 3D texturing techniques volumetrically visualize deformation results in real time [6]. In this case, if a set of vertices of the geometry N is updated into N' by physics-based simulation, our process creates base polygons using new coordinates N', and a set of previous coordinates N is used for their texture coordinates. Since voxel values in each tetrahedral element are correctly mapped on newly generated base polygons through this process, deformed volumetric image is visualized.

In addition to visualize model deformation with image voxels, we focus on coloring volumetric images. We utilize this representation for visualizing selected volume area by the virtual forceps. Also, volumetrically colored elements are effective to visualize internal stress simulated by the proposed FEM model.

Based on the same concept of interactive volume visualization of deformable geometry, we do not update any image voxels. For overlaying color on each element in real-time, we developed the following algorithms.

(1) Slicing tetrahedral grid and obtain the cross sections consisting triangle and square polygons. The cross sections are called based polygons in this paper.
(2) The color of vertices of base polygons is calculated from vertices color of tetrahedral element by simple linear interpolation.
(3) The color is set to the base polygons before texturing voxels.

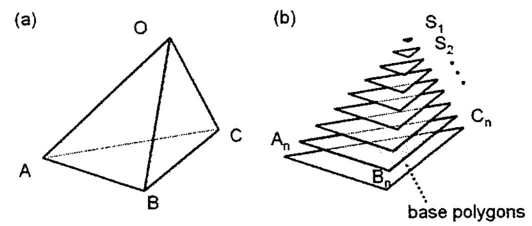

Fig. 5 Real-time color overlay per vertex for tetrahedral slice-based volume rendering

RESULTS

Overall methods were implemented on a standard PC (CPU: Pentium4 3.2GHz, Memory: 2048MB, GPU: nVidia Quadro FX3400) and a prototype volume manipulation system was developed. The PHANToM Omni was implemented into our system as 6DOF input and force feedback device. To confirm applicability and performance of our approach, we tried to simulate pinching manipulation on volumetric lung and liver models.

The lung geometry was semi-automatically created from 256x256x256 16bit CT dataset. Amira 3.1 (Mercury Inc.) was utilized to extract lung region and tetrahedral grid generation. The geometry