Realistic Deformable Models for Simulating the Tongue during Laryngoscopy

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Abstract

During the procedure of Laryngoscopy, an anaesthetist uses a rigid blade to displace and compress the tongue of the patient, and then inserts a tube into the larynx to allow controlled ventilation of the lungs during an operation. This procedure can sometimes be difficult and even life threatening, and there is therefore a need for regular training. Currently, plastic models are used for this purpose, and these have many disadvantages. Computer simulation is an attractive alternative, however, for proper realism it is necessary to build a model of the upper airway. In particular, we need a deformable model that can realistically simulate the behaviour of the tongue as it is compressed by the blade. We start from medical images, extract the details that characterise the subject, and then incorporate these in a finite element model to investigate how the tongue tissue behaves in response to the insertion of the blade, when it is subjected to a variety of loading conditions. The results show that, within a specific set of tongue material parameters, the simulated outcome can be successfully related to the experimental laryngoscopic studies. Further research is underway to apply these results in a virtual reality simulation for laryngoscopic training. One main problem to be solved is computing the deformations in real time.

Key words - deformable models, simulation, tongue, laryngoscopy

1. Introduction

Laryngoscopy is a procedure, carried out by anaesthetists, in which the tongue of a patient is compressed and displaced to one side of the mouth using a rigid blade (Figure 1). A tube is subsequently inserted into the larynx to feed a mixture of oxygen and anaesthetic gas to the lungs. Difficult intubation can cause mortality associated with anaesthesia, but yet the reasons for difficult laryngoscopy have not been completely identified or explained. Anaesthetists must be trained to respond rapidly and correctly to a wide variety of circumstances occurring during laryngoscopy, many of which are critical to the well being of the patient. Training in the use of the laryngoscope is currently carried out using plastic models of the head. These have a number of clear disadvantages. They are not realistic, they offer no variation, they do not attempt to simulate difficult airways, and there is no way of assessing the quality of the trainee. For example, there are not penalties for a trainee who uses excessive force. All of these limitations could be overcome by means of a computer simulation using virtual reality with appropriate haptic feedback.

Figure 1: A laryngoscopy in progress. The anaesthetist is using a probe to measure the extent of the view of the vocal cords.

One of the key research issues in the development of medical simulators is the modeling of soft tissue mechanics and deformation that can represent the variation of its geometry over time with acceptable fidelity. The mechanical properties of the tongue are complex. When deformed, there is a phase, in which the blood is squeezed out of it, followed by a compression phase. Considerable research has been carried out in modeling the tongue, but none the less its properties are still not fully understood. We start from measurements taken from magnetic resonance scans and X-ray films, extract the geometric details, and then incorporate them in our biomechanical models. These models and simulation prototypes proved capable of representing realistic information about the complex mechanisms involved which include the deformable behaviour of the tongue, the geometry of the upper airways, and blade shape.

2. Background

One of the most common and effective approaches to modeling elastic bodies which can undergo finite deformation is to postulate a strain energy function, W, or some of its derivatives with respect to the two independent strain invariants, I_1 and I_2 [Larrabee 1986]. Stress-strain relations applicable to finitely elastomeric polymers have been suggested and investigated by many researchers to describe the deformation behaviour of the stressed soft biological tissues [Sahay 1984].

To formulate a strain-energy function it is necessary first to hypothetise a constitutive law and solve a boundary value problem, and then to design and conduct experiments to test the validity of the theoretical solutions. This finite elasticity approach successfully describes a wide range of biological tissues within the accepted degree of approximation, and the relevance of such an approach has been accepted by several investigators [Fung 1993]. Fung in particular has contributed much to the soft tissue area with interesting experiments that have led to various useful formulations of the strain-energy functions [Fung 1967; Fung 1979]. Many other models have been suggested in the literature [Hart-Smith 1966], [Alexander 1968], [Sahay 1984], and [Maurel et al. 1998]. These strain-energy formulations have generally been validated in uniaxial experiments, and only a few have been applied to biaxial experiments [Tong and Fung 1976; Huang et al. 1990]. However, uniaxial experiments cannot exactly define the mechanical properties of a three-dimensional solid. More specifically, the constitutive equations that characterise soft tissues in three dimensions cannot be generalised from one-dimension [Larrabee 1986]. In general, all these functions have a polynomial or exponential form, and could be applied to any soft tissue type. Actually, considering all the approaches as well as the specific nature and conditions of experiments, it is unlikely that one model may be more suitable or reliable than the others in all circumstances. Some analysis of suitability can be established by comparing the models according to some specific criteria, for example, the number of material constants in the model. It is clear that the larger the number of constants required, the more difficult the model is to validate. Since real material data, from empirical studies on biological soft tissues, is commonly not available, it becomes necessary to maintain a proper balance between the complexity of the model, and a knowledge of the relevant parameters and the constitutive laws that are available to back up the model. Therefore, strain-energy functions with fewer material constants are preferred.

Non-linear viscoelastic models have also been proposed in the context of strain-energy functions. The main point in common between the linear and non-linear viscoelastic theories is that of the memory hypothesis [Christensen 1971]. In terms of a relation between stress and strain, this simply means that the current value of stress is determined not only by the current value of strain, but also by the complete past history of strain. Simo has proposed a fully three-dimensional finite-strain non-linear viscoelastic model, capable of accommodating general anisotropic response and general relaxation times [Simo 1987].

To sum up, the question of the "proper" choice of strain measure has received a great deal of attention in the literature, but no general agreement in the validity of "the" model has been obtained so far. Thus, the formulation of strain-energy functions for biological soft tissue deformation is still a promising area of research. To solve the equations governing deformable models more and more researchers have tended to use finite element methods. Finite element analysis is recognised as the most powerful tool available to obtain quantitative descriptions of the behaviour of soft tissue, given the limited availability of experimental data such descriptions could not be realised otherwise [Lee and Tseng 1982]. The reason the finite element method was chosen as the underlying formulation for the tongue model is that it provides a way of calculating continuum mechanical problems which are characterised by a complicated external and internal geometry.

3. Simulation

In order to reconstruct the geometry of the tongue we used just four lateral magnetic resonance slices. These were the middle lateral image, and three slices spaced at 10mm intervals on the left-hand side. Use was made of the symmetry of the mouth so that the right hand side was the reflection of the left. Each slice was segmented by hand to identify the boundaries of the tongue tissue. To simulate mouth opening around the laryngoscope, a 2.5cm wide plastic device was positioned between the lower and upper incisors of the patient during the scanning session. Once we obtained the tongue data, we converted it into finite elements following the main underlying orientation of the fibres of the tissue to be used in the simulation. In particular, 280 8-noded, isoparametric, three-dimensional bricks were used for representing the tongue structure with an additional ninth node with a single degree of freedom for representing the pressure. This element uses a mixed formulation for incompressible analysis based on the Hermann formulation for representing a three-dimensional arbitrarily distorted cube. The following type of displacement assumption and mapping from x, y, z space (global coordinates) into a cube in the α, β, γ space (local coordinates) is used by this element:

$$x = a_0 + a_1 \alpha + a_2 \beta + a_3 \gamma + a_4 \alpha \beta + a_5 \beta \gamma + a_6 \alpha \gamma + a_7 \alpha \beta \gamma$$

$$u = b_0 + b_1 \alpha + b_2 \beta + b_3 \gamma + b_4 \alpha \beta + b_5 \beta \gamma + b_6 \alpha \gamma + b_7 \alpha \beta \gamma$$

For the assumed strain formulation, the interpolation functions are modified to improve the bending characteristics of the element. The 24 generalized displacements are related to the u – v – w displacements (in global coordinates) at the eight corners of the distorted cube. The stiffness of the element is found by numerical integration using eight points defined in the α, β, γ space. Either the coordinate or function can be expressed in terms of the nodal quantities by the integration functions (similarly for y and z):

$$x = \sum_{i=1}^{8} x_i \phi_i$$

$$\begin{aligned}
\phi_1 &= 1/8 \ (1 - \alpha) \ (1 - \beta) \ (1 - \gamma) \\
\phi_2 &= 1/8 \ (1 + \alpha) \ (1 - \beta) \ (1 - \gamma) \\
\phi_3 &= 1/8 \ (1 + \alpha) \ (1 + \beta) \ (1 - \gamma) \\
\phi_4 &= 1/8 \ (1 - \alpha) \ (1 + \beta) \ (1 - \gamma) \\
\phi_5 &= 1/8 \ (1 - \alpha) \ (1 - \beta) \ (1 + \gamma) \\
\phi_6 &= 1/8 \ (1 + \alpha) \ (1 - \beta) \ (1 + \gamma) \\
\phi_7 &= 1/8 \ (1 + \alpha) \ (1 + \beta) \ (1 + \gamma) \\
\phi_8 &= 1/8 \ (1 - \alpha) \ (1 + \beta) \ (1 + \gamma)
\end{aligned}$$

A surface extraction process was used to reconstruct the three-dimensional shape of the tongue. As to the other components, there are 35 thin shell elements for the blade with data that were obtained from accurate photographs of the laryngoscope with a super-imposed scale to standardize the measurements, and 66 shell elements for the larynx with dimensions that were also extracted from the magnetic resonance scans. Initial values were required for the mechanical properties of the tongue and the laryngoscope blade, and these were set by reference to well established literature [Fung 1993; Timoshenko 1990]. As the simulation proceeded we were able to infer a more precise estimation of the range of possible tongue material values. The *Poisson's ratio* was set to 0.49 in all simulations in line with other studies using volume preserving materials [Chen 1992].

The displacement characteristics caused by blade movements are critical and one important consideration is the space into which the tongue volume can be displaced. This space is bounded by the soft tissue to the side and below, and by the bony parts of the mandible. Our objective was to investigate the material properties and applied force ranges which would give us results comparable with those obtained empirically. Published data of forces applied to the tongue [Bucx 1992], and the peak force values collected from different volunteers in experiments done using the Macintosh blade [Bucx 1992; Bishop 1992; Hastings 1996, McCoy 1995] were used to generate the loading curves. All the authors have described the use of similar methods involving subjects with similar anatomical characteristics, although the peak force varied as much as 40N.

A large number of simulated laryngoscopy runs, with different parameters, were then carried out in which the vocal cords view, obtained by deforming the tongue, was measured against the load applied. A typical simulation is shown in Fig. 2.

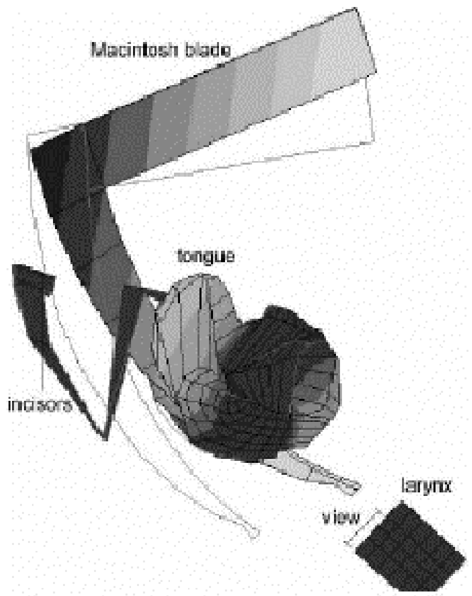

Fig 2. Lateral view of a simulated laryngoscopy using a Macintosh laryngoscope blade. This simulation used the non-linear model with values of peak force = 35N and tongue stiffness = 3*10^6 dyn/cm².

The subject from whom the data was collected was an easy one to intubate in normal conditions. However, with the prosthetic extension to her incisor teeth, the view of the vocal cords was reduced to nothing. Since we knew the exact geometry of the rigid parts of the mouth for both cases, and we can safely assume that in both cases the soft tissues had the same compliance properties, bounds could be put on the plausible ranges of the parameters of the simulation.

Three different types of deformable models were used in the simulation. The first of these was a linear elastic model, where the displacement of each element is proportional to the force applied. This is a simple model to compute, but not likely to capture the complexity of muscle tissue. The second was a non-linear model, where at first it is relatively easy to compress the tongue and the model behaves almost linearly. However, at a particular point, the behaviour of the material becomes stiffer and more difficult to deform, even when applying higher forces. This model is represented by the Mooney-Rivlin formulation [Findley and Lai 1967; Mooney 1940] that not only provides an accurate representation of the mechanical response for large ranges of deformation [Rivlin and Sawyers 1997], but is also simple enough for setting the material parameters (containing only two material constants) for the analyses. This material has a non-linear relation between stress and strain, and thus, is consistent with an anaesthetist own perception when trying to push the tongue. The third approach is to use the previous non-linear model and introduce even more complex behaviour into it by adding viscoelastic constraint functions. As with other viscoelastic materials such as polymers, the tissue of the tongue is characterised by a non-linear rate-dependent viscoelastic model, which can represent a stress relaxation phenomenon. We use a fully three-dimensional non-linear viscoelastic formulation [Simo 1987], capable of accommodating general relaxation times. In practice, the previous pure non-linear model can be said to have an infinite relaxation time, since there is the absence of viscosity.

Hastings et al. [Hastings 1996] addresses the stress relaxation phenomenon during laryngoscopy. Although it is still controversial, and does not provide full evidence for the observed facts, the work offers a more accurate representation of the behaviour of the soft tissues during the blade manoeuvre. The authors report that peak force decreased with time during their laryngoscopic experiments. However, as far as stress relaxation is concerned, there is no clear evidence about the possibility of establishing a baseline for measuring this phenomenon during the short period (in average, less than 17s) in which their laryngoscopic experiments were realised.

To avoid getting overly complicated and speculative, the relaxation data of the viscoelastic model were sampled only for a few trial values. Although they do not give a precise solution, they can suggest a representation that may offer an intermediate solution with a great simplification in the number of analyses required. Our results showed that the non-linear model behaves most closely to the experimental studies, followed next by the viscoelastic model with 11s of relaxation time. However, caution should be exercised in interpreting these results due to the incompleteness of empirical data.

4. Experimental Validation

The three-dimensional finite element model was subjected to an empirical validation that demonstrated its properties (Rodrigues et al 2001). Two different laryngoscopic experiments were carried out on the same subject. In the first experiment, the anatomy of the subject on whom normal rigid laryngoscopy was carried out using a curved laryngoscope blade was used as the basis of our geometric model. In the second experiment, the anatomy of this normal subject was changed empirically by using a prosthesis, 1.5cm thick, attached to her upper incisors, so as to simulate a common difficult laryngoscopic situation, the patient with prominent maxillary incisors. During the procedure, a measurement was made of view of the laryngeal inlet. In the normal laryngoscopy the visible length was estimated as 1.65cm. In the difficult case, simulated by using a maxillary prosthesis, the anaesthetist saw just the back of the larynx. There was no chance to rotate the probe forward to get any angle for maximum forward position.

Our experiments show that it is possible to set up an model of the human upper airway, and replicate both difficult and easy cases of laryngoscopy. For use as a training system, it will be necessary to provide a three dimensional display and appropriate haptic feedback. The view seen by an anaesthetist during an actual laryngoscopy is shown in figure 3. This may be compared to a view calculated during a computer simulation in figure 4.

Clearly, little extra is required in the way of graphics to make the latter useful. However, work does need to be done in accelerating the computation to provide real time performance. This may be achieved either by using special purpose hardware, for example field programmable gate arrays, or by precomputing a number of possible scenarios for conducting a laryngoscopy.

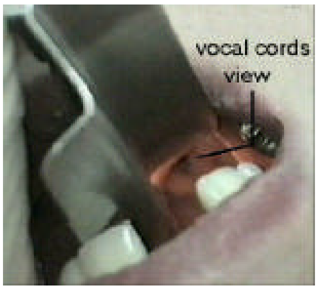

Figure 3: The view of the larynx seen by an anaesthetist during a successful procedure.

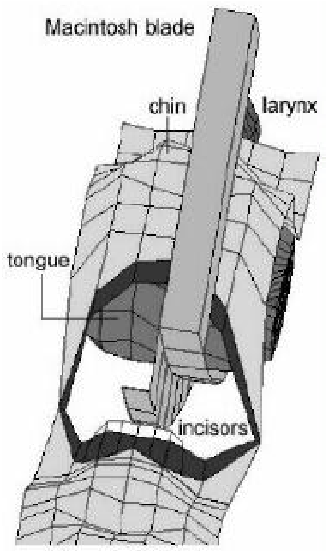

Figure 4: Simulation of the view seen by the anaesthetist during laryngoscopy. The vocal cords are not visible.

5. Conclusions

Some important mechanical aspects of laryngoscopy have been modeled and proved capable of representing realistic information about the complex mechanisms involved. In particular, the non-linear, provided a very good degree of fit between experimental studies and simulated tongue deformations, in both easy and difficult situations. The finite element analyses of deformations (views) and comparisons of different formulations for investigating the biomechanics of the tongue proved useful to define a practical approach for simulating realistic laryngoscopy. We believe that these soft tissue models for simulating the tongue during laryngoscopy have potential to make predictions about the behaviour of the upper airways. Consequently, the models developed in this work may help to predict the outcome results and improve the overall safety of the procedure, as well as have implications for future laryngoscope design.

The laryngoscopy model has been developed, but it has not yet been used systematically nor completely modeled to explore all the mechanical aspects of laryngoscopy. However, we believe that computer models of laryngoscopy may allow training based on the use of large numbers of simulations derived from suitable image data. The advantage is that the novice can be introduced to uncommon conditions that would only arise rarely in clinical practice. Difficult intubation of whatever cause is uncommon enough. The widely varied pattern of abnormality that can end up resulting in difficulty is likely to lead to expanded knowledge and understanding of these conditions as well as suggesting new solutions to problems that eventually can occur. In the short term, a considerable amount of training will be undertaken using training devices rather than patients, and more advanced training and assessment will be conducted using simulators. Consequently, these results can potentially be used to construct a specific laryngoscopic simulator in a non-invasive computerized procedure for training and improvement of diagnosis with fewer complications, lower expense, reduced patient discomfort, and safety in more efficient way than the traditional methods used so far of simulations and lectures.

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