DEFORMABLE MODELING OF FACIAL TISSUE FOR CRANIOFACIAL SURGERY SIMULATION

Invited Paper

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Abstract: While deformable object modeling has been studied in computer graphics for more than two decades, only a few applications in surgical simulation have been developed which provide both real-time and physically realistic modeling of complex, non-linear, tissue deformations. Especially in craniofacial surgery the prediction of soft tissue changes - which are the result of the alteration of the underlying bone structure - is critical to the surgical outcome. Up until today the prediction of these tissue changes, and therefore the prognosis of the postoperative appearance of the patient, is still based on empirical studies of the relationship between bone and tissue movements: There exist no physical model, which takes into account the individual patient anatomy in order to simulate the resulting tissue changes during craniofacial surgery.

In this article we present two different deformable tissue models, which are integrated in an interactive surgical simulation testbed. Both techniques allow the precise preoperative simulation of the resulting soft tissue changes during craniofacial surgery and the visualization of the patient's postoperative appearance. The different deformable models are described in detail and both are applied to the same craniofacial case study. The simulation results are shown and compared with regard to the speed and accuracy of the prediction of the patient's postoperative appearance.

I. INTRODUCTION

Computer-aided surgery is a relatively new field that has made a great impact on medicine in the last few years *[Tay96] [Zon94]*. Particularly surgical simulation has many applications in medical education and training, surgical planning and intra-operative assistance. Compared to traditional methods and considering the high costs of animal specimens, surgical simulation can be used in medical education and training to reduce costs, to provide experience with a greater variety of pathologies, and to enable the trainee to repeat training procedures over and over. In surgical planning, a simulator can reduce costs and save time by replacing stereolithographic models while still providing patient-specific anatomy in order to rehearse difficult procedures. Intra-operatively, computer modeling can help with the navigation of instruments by providing a broader view of the operation field. In combination with robotics it even can supply guidance by predefining the path of a biopsy needle or by preventing the surgical instrument from moving into harmful regions *[Bla97]*.

The most challenging task in this new field is the realistic modeling of the resulting soft tissue changes during surgery. Although deformable object modeling has been studied in computer graphics for more than two decades, only few applications in surgical simulation have been developed which provide both real-time and physically realistic modeling of complex, non-linear, tissue deformations *[Gib98]*. In particular, it is often difficult to obtain accurate tissue properties, which describe the elasticity of living tissue, and it is even harder to conduct those kinds of experiments required to verify the precision of the computer simulations. Therefore, the computer-based simulation of the biomechanical behaviour of human soft tissue has been addressed only in the last few years by specialized research sites *[Bro96a] [Cot96] [Cov93] [Del94] [Gib97] [Koc96] [Pat96] [Pie95] [Tan95]*. Although this approach requires both high-performance computation as well as high-end graphics capabilities, the great advantage of the ability to simulate the elasticity of soft tissue and the functionality of human organs has led to establish a new research field in the area of computer aided surgery: ***deformable modeling in surgical simulation*** *[Sze98]*.

Among other surgical areas in which deformable modeling of soft tissue can lead to a better understanding of the patient's anatomy and may improve the outcome of surgical procedures, craniofacial surgery stands out: Craniofacial surgery has long been an extremely productive application area for the development of computer aided surgical methods - much of the initial work in 3D reconstruction and visualization of patient anatomy has come from this field *[Alt93] [Bil95] [Den88] [Euf94] [Has95] [Hem91] [Lo94] [Van83] [Zei95]*. In craniofacial surgery the goal is not only to improve the functionality, but also to restore an aesthetically pleasing face. Therefore the realistic prediction of the ensuing soft tissue changes is substantial. With traditional methods only a limited forecast of these soft tissue changes can be achieved *[Ste94]*. The prediction is still based on empirical studies of the relationship between bone and tissue movements *[Ath95] [Krü85] [See92]*. Although this relationship can be found for forward and backward shifts of the lower jaw, in operations involving the upper jaw or other displacements of the lower jaw, there is presently no satisfactory method of predicting the resulting soft tissue changes.

To address this challenge we integrated two different deformable tissue models into our surgical simulation system. One interactive more approximating technique – described in section III.4 – that gives the surgeon the ability to realize nearly real-time simulations of the resulting soft tissue changes and to improve his planning process. And one more precise technique – described in section III.5 - which considers the exact physical properties of the facial soft tissue and can be used off-line to verify the chosen surgical procedure. Both techniques are described in detail and are applied to the same case study in order to validate the accuracy of the simulation results.

II. RELATED WORK

Although deformable soft tissue models are widely used in computer graphics - for example in facial animation - these applications have a different emphasis than those applied in surgical simulation: Their intention is to realistically animate facial expressions, not to simulate the exact physical behaviour of human soft tissue *[Che92] [Che95] [Gou89] [Lee95] [Par96] [Ter90] [Ter91] [Wat92] [Wat95]*. There is no consideration of the individual anatomical structure neither are the actual elastic properties of living tissue taken into account. Therefore these techniques can only be used as a brief guideline in order to develop new deformable tissue models suitable for surgical simulation. However, in the last few years various deformable models have been evolved in order to predict intra- and postoperative modifications - for instance in minimal invasive and orthopedic surgery *[Cov93] [Fis96] [Gib97]*. Beside these surgical areas most of todays leading techniques in deformable tissue modeling have been projected in plastic- and craniofacial surgery.

- A *Craniofacial Surgery Simulation Testbed* was developed by Delingette et al. in 1994, using deformable surfaces and volumes called *Simplex Meshes* in order to simulate surface changes such as those exerted on the facial skin by muscle action *[Del94]*. But this approach does not consider the exact anatomical structure of the soft tissue; instead the user has to specify virtual *Muscles* which then define the connection between the facial skin and the underlying bone.

- To simulate facial palsy and cleft lip surgery, Tanaka et al. developed a deformable tissue model based on a facial muscle model *[Tan95]*. This muscle model is limited to only the mouth and does not take into account the underlying bone structure. It can not be used to determine soft tissue changes resulting from bone displacements as they occur in craniofacial surgery.

- *CAPS*, a computer-aided plastic surgery system developed by Pieper et al., can be used to predict soft tissue changes on individual skin data *[Pie91] [Pie92] [Pie95]*. But this approach does also not take into account the underlying bone structure of the patient and can not be used to predict soft tissue changes affected by surgical bone realignment.

- Koch et al. presented an approach to model soft tissue changes shaped by the realignment of the underlying bone structure *[Koc96]*. In this approach mass-spring and finite-element methods are used in order to model the elasticity of the facial tissue. But their methods are not integrated into a surgical simulation system; instead commercial available software packages like *AVS* and *Alias* are used to build the necessarily three-dimensional surface models *[Adv98] [Ali98]*. Because of these time consuming steps this approach has only be tested on the *Visible Human Project* dataset and has not been applied to actual patient data *[Nat98]*.

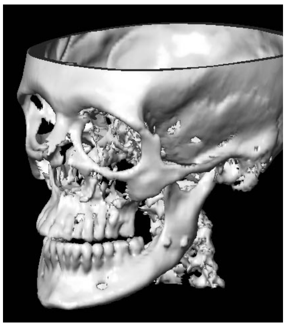
Although the discussed methods are very interesting and auspicious, each of them only allows the simulation of soft tissue changes in a very specific manner and none of them is yet integrated into a surgical simulation system which empowers both real-time and physically realistic modeling of complex, non-linear, tissue deformations: Therefore none of them have yet been applied to actual clinical cases.

III. MATERIALS AND METHODS

The goal of the presented craniofacial surgery simulation system is to enable precise and interactive manipulations of the three-dimensional reconstruction of the patient's bone and to predict the resulting soft tissue changes in order to visualize the patient's postoperative appearance. As stated, the focus of this paper is the deformable modeling of facial tissue. Therefore, the next two sections elucidate only briefly the general setup of our system, the data acquisition and visualization and the simulation of the bone realignment (for a detailed description the reader is referred to *[Kee96a]*). Section III.4 describes in detail the mass-spring and section III.5 the finite-element tissue model. Both models are integrated into the surgery simulation system which is implemented in *C++* on Silicon Graphics workstations, using the object-oriented graphics library *Open Inventor [Wer94]*. Each algorithm has its own *Motif* dialogue window, which gives the user the flexibility to easily tailor parameters to suit the needs of their application.

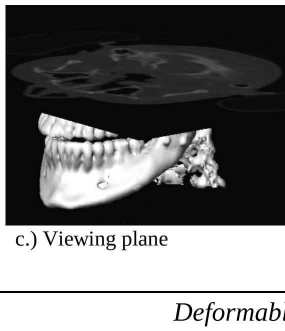
III.1 Data Acquisition and Visualization

The *Marching Cubes* method is used to reconstruct the skull from computer tomography data *[Lor87]*. This three-dimensional skull reconstruction provides the basis for the interactive surgical simulation of the bone realignment. A Cyberware scanner obtains the geometry of the patient's skin surface *[Cyb98]*. After data acquisition, and 3D surface reconstruction an adaptive reduction technique is used to decrease the size of both datasets *[Kee97]*. This compression algorithm reduces the amount of triangles used to represent complex anatomical structures up to 80 % without sacrificing the visible details of the objects. A semi-automatic method registers the skin and skull data, which are obtained in different coordinate systems. The user provides corresponding anatomical points in both datasets. A matrix is calculated by least squares fitting, which transforms the data from one coordinate system to the other. Figure 1a shows the *Marching Cubes* reconstruction of the patient's skull. As shown in Figure 1b, a 3D *Cephalometry* tool allows the physician to easily measure the exact anatomical structure of the patient's skull and skin. Figures 1c and 1d illustrate the *Viewing Plane* tool that can be used to generate a section at an arbitrary angle and visualize particular structures of interest - a volume-rendered image is used as a reference for a more detailed structure localization. The system also supports visualization of the scene in stereo. As shown in Figure 1e, the *Implant Carving* tool can be used to model a custom prosthesis with a desired shape. To verify the tissue model, a *Verification* tool as shown in Figure 1f, is used to visualize the shape differences in the simulation with respect to the actual postoperative finding.

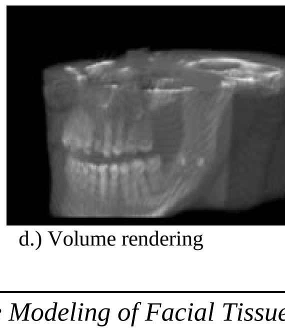


a.) Marching cubes recon准onstr.

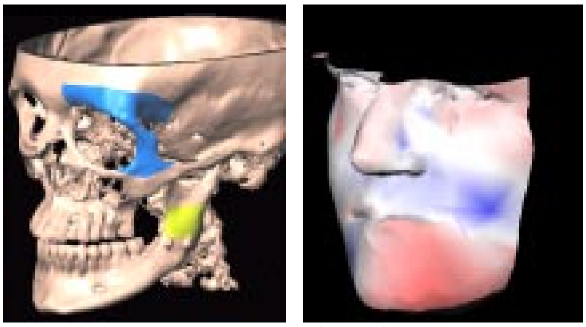
b.) 3D cephalometry



c.) Viewing plane



d.) Volume rendering

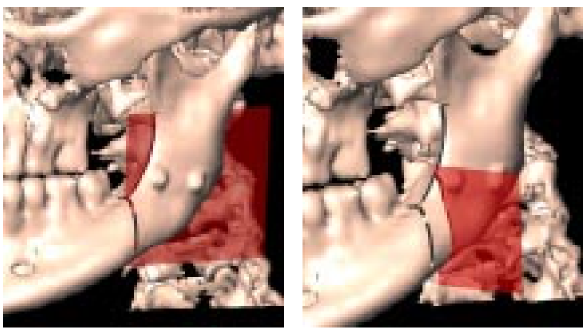


e.) Implant carving f.) Verification

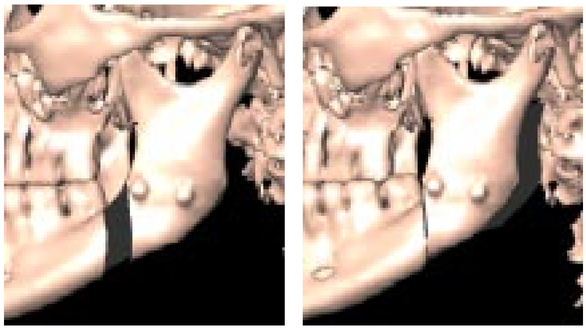
Figure 1: Data acquisition and visualization

III.2 Simulation of Bone Realignment

In order to perform the surgical simulation, first the bone realignments are alleged. Afterwards, the ensuing tissue changes are calculated with respect to these bone realignments. The tissue models used in these calculations are described in detail in section III.4 and III.5. In contrast to other systems that allow only predetermined surgical procedures, our system provides the flexibility to apply any craniofacial operation on the bone structure and manipulate it at will. These changes are made with a *Cutting Plane* with which the bone is split in half. After the split, each bone segment is defined as a unique object that can be further manipulated. The *Cutting Plane* can be changed in shape and orientation to enable the simulation of different procedures. Figure 2 shows the simulation of a Dal-Pont Osteotomy, which splits the ascending branches of the mandible *[Dal61]*. The interactive simulation is analogous to a real operation. To the lingual (inner) side of the ascending branch a horizontal cut is applied which divides the lower jaw as far as the middle. On the vestibular (outer) side a vertical cut is performed half way through the bone. Then a sagittal cut splits the bone lengthwise in half. With an object manipulation tool, the now separated frontal mandible is pushed backwards to correctly align it with the maxilla. These steps can be accomplished on a common graphics workstation in approximately 10 minutes.



a.) Splitting by Dal-Pont b.) Cutting bone



c.) Removal of bone d.) Bone realignment

Figure 2: Simulation of bone realignment

III.3 Soft Tissue Properties

In order to build accurate deformable soft tissue models the elastic properties of living tissue have to be studied and well understood. There exist numerous articles and books, which describe the mechanical and functional properties of the different anatomical structures: The most fundamental one was written in 1993 by Y. Fung *[Fun93]*. In this section we give a short overview of those features, that effect the deformable tissue models.

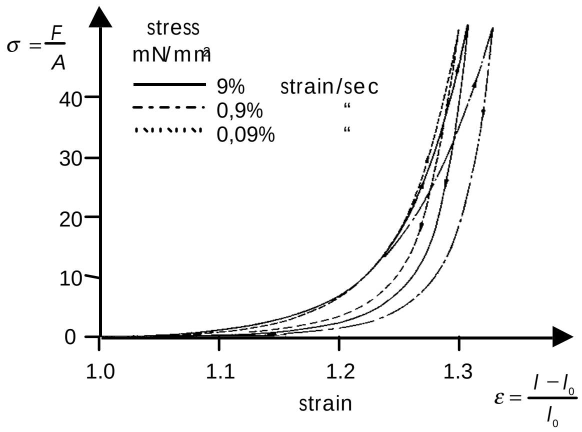
Facial skin is built by three layers of tissue: epidermis, dermis and hypodermis. The epidermis, the outer tissue layer, is just 0.1 mm thick and is supported by the underlying dermis layer, which much thicker (0.6 mm to 3.5 mm) is responsible for the elasticity of the skin. The underneath hypodermis layer contains the fatty tissue and only slightly affects the elasticity of the facial skin. The connection between these skin layers and the underlying bone structure is made by the facial muscles, which are built by actinfibers and do have an important impact on the facial elasticity. Therefore, the biomechanical behaviour of the facial soft tissue is based on the following three substances:

- Actin; muscles are built by this protein, fibers of which have an almost linear strain-stress relationship.

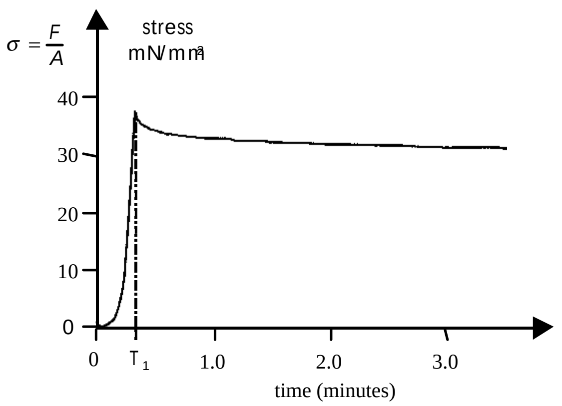
- Elastin; most of the facial tissue fibers contain this protein, which is responsible for the elasticity of the skin. The production of this protein decreases with age. Elastin fibers have a linear strain-stress relationship.

- Collagen; this protein is the main component of the facial soft tissue. Various different collagen fibers with diverse mechanical properties exist. Therefore, the elasticity of facial soft tissue is mainly determined by the mechanical properties of these collagen fibers, which in general have a non-linear, visco-elastic, strain-stress relationship.

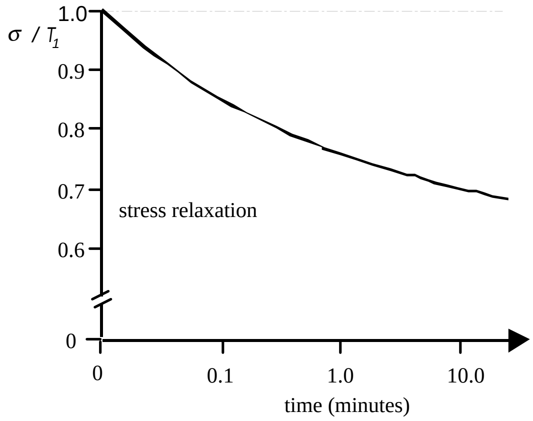
Thus, the facial soft tissue comprises various substances with different elastic properties, which can not be modeled separately. Rather the biomechanical behaviour of the whole facial skin has to be determined by experimental studies. One of these experiments is the measurement of the uni-axial strain-stress relationship, which is shown in Figure 3a. It should be noted that there are short-term as well as long-term relaxations, which means that the force necessary to maintain a certain strain decreases with time (see Figures 3b and c). Furthermore there exists a certain pre-stress along so called Langer-lines, which in this study was not taken into consideration.



a.) Strain-stress relationship



b.) Short-term relaxation



c.) Long-term relaxation

Figure 3: Soft tissue properties

To model these elastic tissue properties two main approaches can be found:

- Virtual reality deformable models: Most of these techniques are based on mass-spring models, with which linear elastic properties can be simulated in real-time *[Bro96a] [Bro96b] [Lee95] [Pro95] [Ter90] [Ter91] [Wat92]*. Therefore these models are used only to realistically animate tissue deformations, not to simulate the exact physical behaviour of human soft tissue.

- Mathematical deformable models: These classical engineering techniques – like the finite-element method – allow the exact mathematical description of non-linear, anisotropic and visco-elastic material properties *[Bel90] [Che92] [Che95] [Gou89] [Hem91] [Hol95] [Lar86a] [Lar86b] [Lar86c] [Pie92] [Pie95] [Sag94]*. However, these techniques are very CPU intensive and the tissue deformations can not be simulated interactively.

To benefit from the advantages of both approaches, we integrated a deformable mass-spring tissue model – which is described in detail in section III.4 - as well as a deformable finite-element tissue model – specified in section III.5 - into our craniofacial surgical simulation system.

III.4 Deformable Mass-Spring Tissue Model

A layered mass spring tissue model for facial animation was first introduced by Waters in 1992 *[Wat92]*. In order to use this model for surgical simulation, it has been modified to take individual patient skin and bone structures into account. For each triangle of the 3D reconstruction of the facial skin one basic tissue element as shown in Figure 4 is created.

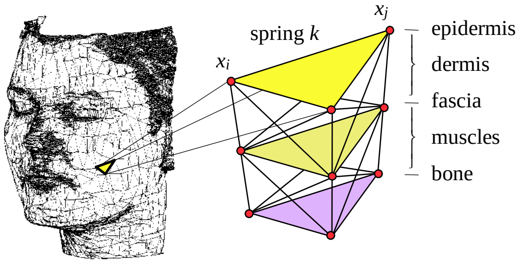
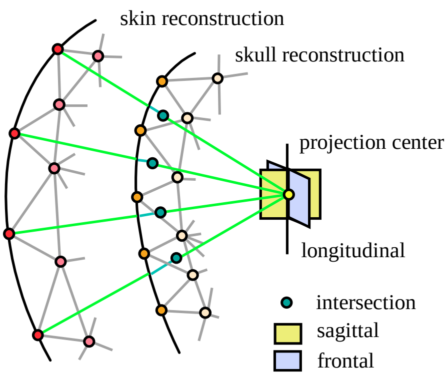
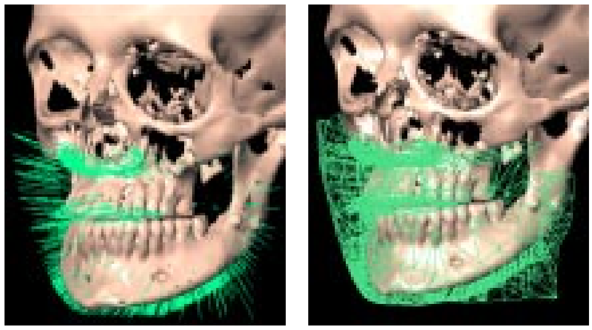


Figure 4: Basic element of the mass-spring model

The basic tissue elements are generated by connecting each skin vertex to the underlying skull structure. As shown in Figure 5a, b and c, we project all skin vertices towards a user-specified point - e.g. the center of gravity of the patient's data. The first intersection with the skull provides the corresponding point for each skin vertex. If no intersection is found, a corresponding point is automatically generated by interpolating the distances between the skin and the skull vertices of the adjacent elements. All found intersection points are then integrated into the skull data in order to apply the simulated bone realignment to the tissue model.



a.) Projection of skin vertices



b.) Skin-bone connections c.) Mass-spring mesh

Figure 5: Model generation

The mechanical properties of each tissue layer are represented by the various spring constants. To approximate the non-linear strain-stress relationship of the soft tissue each spring constant is defined bi-phasic (see Figure 6). Please notice, that in order to simulate a homogeneous elasticity in each layer the spring constants depend not only on the flexibility of the actual individual tissue, but also on the length of the springs and the size of the basic tissue elements, which is largely determined by the local curvature of the skin since the reduction algorithm – described in section III.1 – generates large triangles in regions of low curvature.

The mathematical formulation of the mass-spring model is described by Newton's motion equation (1). For each node $\mathbf{x}_i$, a differential equation of second degree can be given that depends on the nodes mass m_i, the damping γ_i and the sum $\mathbf{s}_i$ of all spring forces $\mathbf{f}_k$ affecting it. Each spring force $\mathbf{f}_k$ depends on the spring constant c_k as well as on the spring length l_k and their expansion. To model the incompressibility of the soft tissue, a volume preservation force $\mathbf{q}_i$ was added to each node. To solve the given motion equations they are numerically integrated over time. The simulation is finished when all nodes meet minimal velocity and minimal acceleration criteria. For a detailed mathematical description of the mass-spring model the reader is referred to *[Kee96c]*.

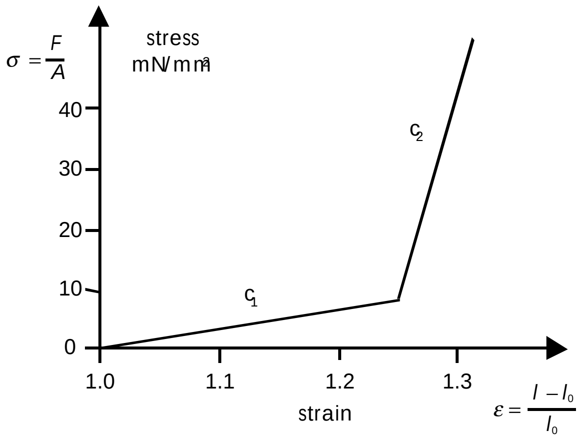
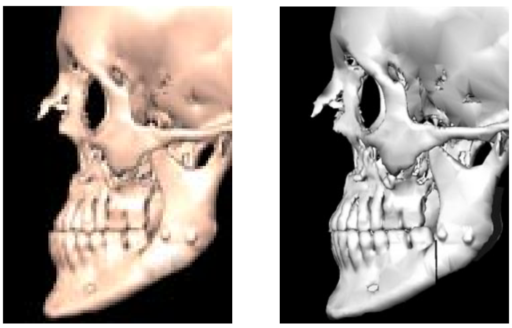


Figure 6: Bi-phasic spring constants

$$m_i \frac{d^2\mathbf{x}_i}{dt^2} + \gamma_i \frac{d\mathbf{x}_i}{dt} + \mathbf{s}_i = 0 \quad (1)$$

$$\mathbf{s}_i = \sum_k c_k \left(1 - \frac{l_k}{\left|\mathbf{x}_j - \mathbf{x}_i\right|}\right)\left(\mathbf{x}_j - \mathbf{x}_i\right) + \mathbf{q}_i \quad (2)$$

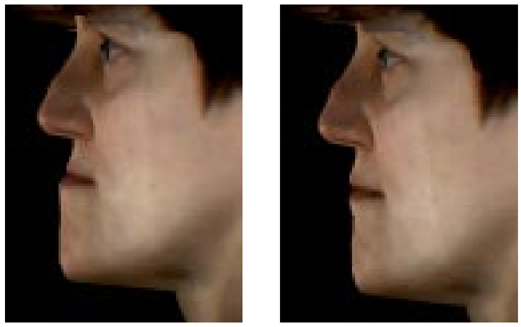
To verify our approach the mass-spring model as well as the finite-element model – described in section III.5 – are applied to the same case study, in which the patient undergoes a Dal-Pont Osteotomy and her lower jaw is moved posterior (see Figure 7)



a.) Preoperative bone data b.) Bone realignment

Figure 7: Bone structure of the case study

Figure 8a shows the patient's preoperative skin data and Figure 8b the outcome of our study; the simulated postoperative appearance of the patient. Considering 3,080 basic tissue elements, this simulation takes approximately one minute using a SGI High Impact workstation. The actual post-operative findings are given for purposes of comparison in Figure 12.



a.) Preoperative data b.) Simulation results

Figure 8: Mass-spring simulation

III.5 Deformable Finite-Element Tissue Model

The basic idea of the finite-element method is that a continuum is approximated by dividing it into a mesh of discrete elements *[Bat82]*. Different element types are appropriate for different applications; in our approach we use a grid of six node prisms, which are very similar to the basic tissue elements used in our mass-spring model. The prism elements are defined by their corner nodes as shown in Figure 9.

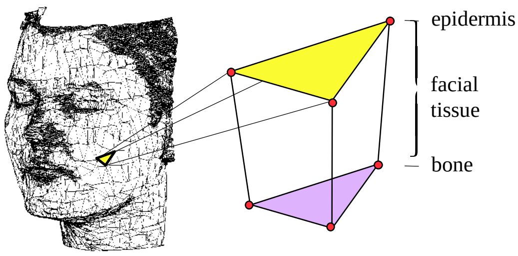


Figure 9: Basic element of the finite-element model

In our displacement-based finite-element model, given displacements are specified for certain nodes – for instance through the simulated bone realignment. These displacements cause strain, which - in a linear-elastic case - is related to stress through Hooke's law, which in turn creates internal forces. In order to bring these forces into equilibrium, a system of differential equations is solved delivering the displacements of the unconstrained nodes.

As shown in Figure 3a the strain-stress relationship of living tissue is highly non-linear and also indicates visco-elastic effects, which means that the history of strain effects the stress; please notice the hysteresis in the loading and unloading process. Each branch of the strain-stress curve can be mathematically described by one non-linear function. However, since the differences are insignificant and because this very much simplifies our computation, we describe the elastic properties of the soft tissue by only one non-linear elastic function – see Figure 10. This technique is known as pseudo-elasticity.

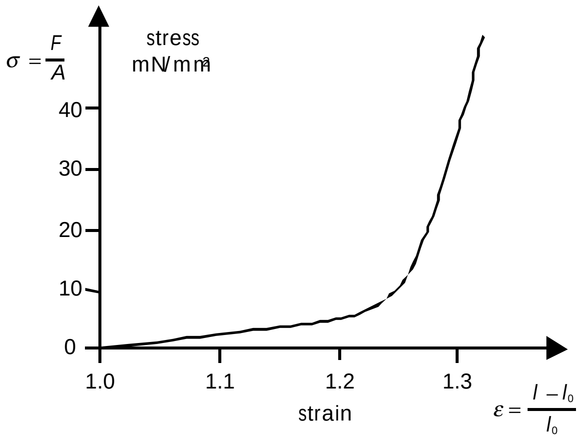


Figure 10: Pseudo-elasticity

In order to model this pseudo-elastic behaviour we use the principal of virtual displacements described by the total Lagrange formulation *[Bat82]*. Using the second Piola-Kirchhoff stress tensor and the Green-Lagrange strain tensor, a linear approximation of the total Lagrange formulation can be stated and solved with a modified Newton-iteration. For a detailed mathematical description the reader is referred to *[Kee96b]*. The finite element formulation for this modified Newton-iteration is given as

$$\left({}^{t}_{0}K_{L}+{}^{t}_{0}K_{NL}\right)\Delta U^{(i)}={}^{t+\Delta t}R-{}^{t+\Delta t}_{0}F^{(i-1)} \qquad (3)$$

where $\boldsymbol{K}_L$ and $\boldsymbol{K}_{NL}$ are the incremental stiffness matrices for linear and non-linear strain, while vector $\Delta\boldsymbol{U}^{(i)}$ represents the incremental nodal displacements at iteration i. $\boldsymbol{R}$ denotes the exterior nodal loads and $\boldsymbol{F}^{(i-1)}$ the nodal forces at iteration $i-1$ which correspond to the element stresses. Equation (3) can be solved using a strain energy function W, which provides the following relationship between stress and strain

$${}^{t}_{0}S_{ij}=\frac{\partial W}{\partial\,{}^{t}_{0}\varepsilon_{ij}} \qquad (4)$$

with $\boldsymbol{S}_{ij}$ as the second Piola-Kirchhoff stress tensor and $\boldsymbol{\varepsilon}_{ij}$ as the Green-Lagrange strain tensor. The strain energy function W for our approach is given by Y. Fung *[Fun93]*. The implementation of this tissue model is based on the object-oriented finite-element library Diffpack *[Lan94]*.

As an example, we used the same patient data as in the mass-spring approach. This provided 3,080 six-node prism elements to the finite-element tissue model. We again simulated an operation of cutting the frontal mandible and pushing it backwards. Figure 11a and b show the patient's skin before surgery and the finite element simulation. It took one minute to set up the initial configuration using a SGI High Impact workstation. The finite element calculations were finished in about 10 minutes.

a.) Preoperative data b.) Simulation results

Figure 11: Finite-element simulation

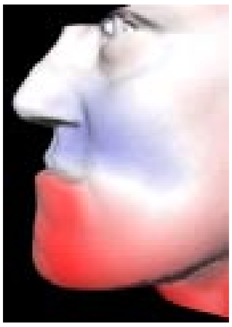
IV. RESULTS

To examine our deformable models we used a case study where the mandible was split, moved backward and pushed to the right. To verify the simulation results the actual postoperative finding is displayed in Figure 12.

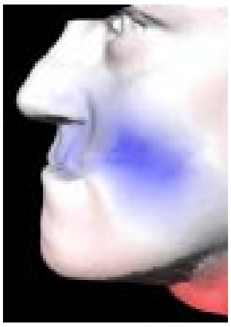
Figure 12: Actual postoperative outcome

Although there is no real objective method to measure the accuracy of our simulations, we color-coded the distances between the pre- and postoperative skin data as well as the distances between our simulation results and the actual postoperative finding (see Figure 13).

a.) Color-coded distances



b.) Pre- vs. postoperative data



c.) Simulation vs. post-operative data

Figure 13: Verification

The comparison shows that in the area of the performed bone realignment – lower jaw – our simulation almost completely covers all actual tissue changes. The inaccuracies in the area of the upper jaw and the neck are mostly due to registration errors and changes in the patient's position during data acquisition.

V. DISCUSSION

The mass-spring approach can not model the exact physical properties of human soft tissue; instead they have to be expressed with tools of masses and springs. However, the presented results show how realistic this technique predicts the tissue changes according to the simulated bone realignments. The achieved results let assume, that in the near future these simulations can be performed in real-time. This will give the surgeon the ability to realize interactive simulations of the resulting tissue changes and to improve his planning process. If one simulation does not yield his expectations, he is able to undo it and try another operation. On the other hand, the finite-element approach models the physical properties of the facial soft tissue more precisely. Although the simulation is time consuming and cannot be done interactively, it highly improves the precision of the simulation and can be used off-line to verify the chosen surgical procedure.

VI. CONCLUSIONS

The presented deformable tissue models both allow the simulation of soft tissue changes during craniofacial surgery and the prediction of the postoperative appearance of the patient. Both techniques are integrated into a surgical simulation system, which gives the surgeon the ability to work interactively with the patient skull data and to simulate different surgical procedures. The performed bone realignments then can be transferred to the soft tissue through each of the presented tissue models in order to predict the resulting shape differences. The presented case study demonstrates the efficiency and strengths of these new methods. The system has already been used for a few case studies at the Department of Oral and Maxillofacial Surgery at our university and has aided the preoperative planning procedure. We plan to install the surgical simulation system there and clinically verify the implemented techniques.

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