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ABSTRACT

The accurate prediction of the post-surgical facial shape is of paramount importance for surgical planning in facial surgery. In this paper we present a framework for facial surgery simulation which is based on volumetric finite element modeling. We contrast conventional procedures for surgical planning against our system by accompanying a patient during the entire process of planning, medical treatment and simulation. In various preprocessing steps a 3D physically based facial model is reconstructed from CT and laser range scans. All geometric and topological changes are modeled interactively using Alias.™ Applying fully 3D volumetric elasticity allows us to represent important volumetric effects such as incompressibility in a natural and physically accurate way. For computational efficiency, we devised a novel set of prismatic shape functions featuring a globally $C^1$-continuous surface in combination with a $C^0$ interior. Not only is it numerically accurate, but this construction enables us to compute smooth and visually appealing facial shapes. An extended evaluation and quantitative analysis of a clinical test series with several female and male patients clearly demonstrates the performance of our framework.

CR Descriptors: I.3.5 [Computational Geometry and Object Modeling]: Physically Based Modeling; I.3.7 [Three-Dimensional Graphics and Realism]; I.6.3 [Applications]; I.6.4 [Model Validation and Analysis]; I.6.5 [Model Development] Modeling methodologies

Additional Keywords and Phrases: Finite Element Method, Facial Surgery Simulation, Facial Modeling, Data Reconstruction.

1 INTRODUCTION

1.1 Background

There is a wide spectrum of facial malformations and diseases which maxillofacial and craniofacial surgeons have to take care of. This includes but is not restricted to diseases like injuries and tumors as well as deformities due to inherited syndromes (e.g. Crouzon-Syndrome, Apert-Syndrome) or developmental disorders (e.g. disturbed growth of the jaw). For patients seeking for surgical treatment in order to correct such malformations it would be of high benefit to have a means to predict the post-surgical appearance of their face in a reliable way. Further, facial surgery has to strive for the reconstruction of a balanced face as even very fine variations of facial proportions can affect the appearance of a face strongly and thus distort its harmony [10].

The prediction and planning of surgical procedures to correct such aberrant skeletal anatomy can to date only be performed in a two-dimensional way from one single perspective, most often the profile view of the patient, such as illustrated in the top row of figure 2. All other views only can be estimated roughly. Hence, the fullness of the lips, the width of the nose, the width and projection of the cheekbones and the influence of the surgery on the appearance of the eyes cannot be predicted. For the time being, only the surgeon’s experience gives information on the overall outcome of the patients facial appearance (see e.g. [7, 3]).

Therefore, both surgeon and patient have a strong need for a method which enables them to compute highly realistic pictures of the expected post-surgical shape during the planning of a surgical procedure.

1.2 Previous Work

The field of facial modeling has been an area of growing research efforts for more than a decade. First approaches such as [19] were based on geometric deformations using parametric

Figure 1: Illustration of the physically based model:
(a) model of our example patient
(b) pre-surgical situation of example patient
surfaces and aimed primarily at facial animation. Later, physically based simulation paradigms were adopted in order to model more accurately the physical properties of elastic materials (see e.g. [22]). For a survey of facial animation see e.g. [15].

Back in 1986, Larrabee [11] stated a finite element model of skin deformation. This work was followed by Deng’s Ph.D thesis [5], where she presented an analysis of plastic surgery by means of the finite element method. In 1991, Pieper [17] summed up his efforts to provide a system for computer-aided plastic surgery in his Ph.D thesis. To our knowledge, this is the first time that facial simulation in combination with finite element modeling was employed as a means of planning surgical procedures. He focussed on plastic surgery and therefore concentrated on cutting and stretching of skin and epidermis rather than on repositioning bones. Further, his model lacked the resolution required for a reliable simulation of very subtle changes in the appearance of a face and did not provide a \( C^1 \)-continuous surface.

Lee et al. [17] presented a promising approach to facial animation where they introduced a layered tissue model based on mass-springs connected to form prism-shaped elements. The facial model is adapted from a template face, takes into account various anatomical aspects and aims at facial animation.

Koch et al. [15] proposed a method which provides a \( C^1 \)-continuous finite element surface connected to the skull by springs. This model is generated directly from individual facial data sets and has successfully been tested for surgery simulation and emotion editing [12] on the Visible Human Data Set. Although providing very promising results the model lacks true volumetric physics.

Therefore, in the field of surgery simulation, attention focussed on the development of volume-based models in combination with more and more sophisticated finite element solution schemes. Keeve et al. [13] presented a system for facial surgery simulation combining Lee’s layered tissue model with a finite element approach in order to solve the inherent partial differential equations. Another interesting approach aiming at real-time applications was proposed by Bro-Nielsen et al. [4]. Both methods only made use of linear interpolation within the elements and therefore suffer from \( C^0 \) artifacts both on the surface and within the volume.

Recently, Roth et al. [19] presented a versatile framework for the finite element simulation of soft tissue using tetrahedral Bernstein-Bézier elements. They incorporated higher order interpolation as well as incompressible and nonlinear material behavior, but again restricted themselves to \( C^0 \)-continuous interpolation across element boundaries.

Apart from the limitations mentioned above, all previous approaches lack an elaborate validation and error analysis with respect to craniofacial surgery.

### 1.3 Our Approach

To optimize both accuracy and rendering quality our goal was to combine the physical correctness of volumetric finite element simulation with the superior quality of the \( C^1 \)-continuous surface of [5, 15]. Furthermore, a validation of the proposed model will investigate its applicability to facial surgery simulation.

As a first major contribution we therefore extended the surface-based approach of [15] to volumetric physics which involved both the reformulation of the mathematical and physical foundations and the redesign of the special purpose triangular finite element model of the human face. As a result we devised prism shaped volumetric elements an illustration of which is given in figure 10a. While still providing a \( C^1 \)-continuous surface using a reduced number of degrees of freedom this volumetric model features a more accurate simulation of tissue behavior including volume preservation and pressure calculations.

The second contribution is a thorough evaluation of our approach with a group of test patients. To achieve this, a prototype application was built which is based on data available from individual patients in a clinical environment. This includes CT scans of the pre- and post-surgical situation in combination with high resolution laser range surface scans. Both individual parameters, such as different stiffness and elasticity of tissue, and the exact bone movements introduced by the surgeon are taken into account in order to simulate as accurately as possible the outcome of real surgery. With validation in mind we re-simulate the surgical procedures carried out on the test group of patients with cranio-maxillofacial abnormalities. The simulation of bone movements is accomplished by means of the commercially available modeling system Alias.³⁴

The outline of the paper is as follows: throughout the report we will accompany a patient who underwent both the conventional and the envisioned treatment before and after the surgical procedure. Section 1.4 describes the process of medical treatment beginning with the first consultative meeting of patient and surgeon. We then compare the conventional and new planning sequence of the surgical procedure. After surgery, the treatment ends with several follow-up checks. An overview of the model build-up and simulation is given in section Section 2. Section 3 discusses the finite element approach including a brief review of the physical foundations and the design of the shape functions. A quantitative evaluation of our method applied to the test group of patients is given in section Section 4.

### 1.4 Accompanying a Patient

The patient accompanied through the following sections is depicted in the right part of figure 1. He suffers from a so-called short face with a deep bite caused by the retropositioned mandible, i.e. the lower jaw, as well a reduced vertical facial height due to a maxilla positioned to high. The left part of the figure illustrates the corresponding physically based model the construction of which will be described in section Section 2.

We will compare the different steps in data acquisition as well as differences in the planning set-up and post-surgical treatment. A general view of the major phases of treatment including timing information can be found in figure 2. The top row contrasts the conventional to the envisioned setting presented in the bottom row.

In both cases the treatment begins with a consulting phase during which patient and surgeon discuss the details of the treatment. The decision, if orthodontic correction of the teeth is necessary has to be made during this phase. In case of a surgical procedure, considerations about the required bone movements and expected convalescence time follow. Afterwards, as a first step of the planning and preparation of surgery, the data acquisition follows. In both cases this comprises lateral and frontal photographs of the pre-surgical situation and the acquisition of a CT scan, if needed. The volume data set can be used to build 3D hardcopies of bone structures on the one hand, and serves as an integral component of the facial model used to simulate the surgery on the other hand (see section Section 2). In addition to the data mentioned above the envisioned planning approach requires
highly accurate surface laser range scans which are also incorpo-
rated into the model underlying the finite element simulation 
onlined in section Section 3.

In the conventional setting, the planning mainly consists of 
structured analyzing lateral X-Ray images upon which studies of 
the actual and envisioned profile can be sketched. These 
sketches in combination with the surgeon’s experience allow the 
estimation of bone movements and cuts necessary to correct the 
disfigurement. Corresponding linear displacements will now be 
transferred to plaster-cast models of the maxillomandibular 
complex (lower and upper jaw) for surgery simulation.

In the envisioned setting, the data acquisition phase is fol-
lowed by the model build-up which consists of skull and surface 
extraction, tissue segmentation, and mesh generation. This pro-
cess and the modeling of the surgical procedure on this model 
will be described in detail in section Section 2.

A convalescence and recovery of four to six weeks and a fol-
low-up time of up to six or even nine months follows the opera-
tive treatment. In addition to the medical controlling of the 
healing process in several follow-up checks, laser range scans of 
the post-surgical facial surface enable to compare the correspon-
dence and the real outcome of surgery. Furthermore, the process of detumesence (diminution of swelling) can be 
documented to a new level of accuracy.

2 MODELING OF TISSUE AND SURGICAL PROCEDURE

In this section, both the construction of the facial model and the 
simulation of the surgical procedure are outlined. We focus on 
the data and algorithms used to build up the model as well as on 
the definition of bone movements which correspond to the actual 
surgery.

2.1 Model build-up

The model underlying to the finite element approach must be 
built upon data available from individual patients including CT 
and laser range scans (LR). After the acquisition of data, the fol-
lowing steps must be performed in order to arrive at a facial 
model capable of capturing the basic properties of tissue and 
offering the instrumentation for surgical simulation at sufficient 
accuracy.

Registration of Volume and Surface Data

In a first step, volume (CT) and surface (LR) data must be regis-
tered in a common coordinate frame. To achieve this we extract 
the facial surface in the CT scan using the marching cubes algo-

rithm of Lorensen et al. [14]. The problem now reduces to find-
ing a transformation which maps the LR surface onto the surface 
originating form the isosurface extraction. Due to the fact that 
the scaling of each surface is determined by the scanning 
method, we are left with finding only a suitable translation and 
rotation in order to register the geometries. To this aim, we 
employ two methods, one of which is fully manual whereas the 
other one is semi-automatic.

We start by manually setting landmark pairs on both surfaces. 
These landmarks are selected to represent characteristic and 
easy to locate facial features, as e.g. the corners of the eyes or 
the mouth and the tip of the nose. An illustration of typical land-
marks is given in figure 3.

Figure 3: Landmarks required for the registration of the laser 
range surface (a) and CT scan (b). The facial surface (b) results 
from a marching cubes isosurface extraction in the CT data set.

Figure 2: Time-line of a patient from first consultations to recovery. The top row represents the conventional sequence and is contrasted 
against the envisioned virtual surgery planning framework shown in the bottom row.
parameters are manually determined by the user aided by the system in terms of cross section images. These result from the intersection of the geometries with an interactively positioned plane scaled to the size of the data sets. Figure 4 shows the geometries, the intersection plane and the cross sections for a rotation around the X axis.

In the semi-automatic set-up, the manual approach serves as a preprocessing step in order to avoid local minima and to accelerate subsequent computations. Starting from in such a manner predefined geometries the registration process can be regarded as the minimization of a scalar error function $E$ defined by the square distances of $n$ corresponding landmark points $u_i$ and $u'_i$. $E$ depends on the rotation $r$ around the three axes as well as on the translation vector $t$.

$$E(r, t) = \sum_{i=1}^{n} (Mu_i - u'_i)(Mu_i - u'_i)^T,$$

(1)

For minimizing $E(r, t)$ we employ a method of conjugate directions. It is a slightly modified version of Powell’s algorithm which is presented in [18].

**Mesh Generation**

The next step consists of the mesh generation for the finite element engine. For this purpose we decimate the LR surface mesh using the approach of Schroeder et al. [20] in combination with local Delaunay re-triangulations. Figure 3a shows the resulting facial surface made up of approximately 30k triangles.

After transforming the reduced mesh according to the matrix computed in the registration step, the facial tissue is tiled with prism shaped elements. We follow the approach of Waters [22] which in essence performs a cylindrical projection of every vertex of the decimated mesh onto the skull inside the CT data set. Figure 5 schematically illustrates the procedure.

**Assignment of Material Parameters**

After the mesh generation we have to assign material parameters in accordance with the CT data to the elements. This is accomplished by a segmentation of CT data into four distinct regions: skin, fat, muscle, and bones. In order to achieve this segmentation we isolate four training regions which are known to belong to one of the above categories. The averages of their CT values provide centroids for further segmentation. In addition, each tissue type is assigned both a Young’s modulus $E$, defining its elasticity and a Poisson’s ratio $\nu$ $0 \leq \nu \leq 0.5$ which describes its compressibility. The assignment of such values is done according to [11]. Linear interpolation between the centroids provides us with intermediate values both for $E$ and $\nu$ for all voxels of the CT data set. Figure 6 shows a schematic overview of the procedure.
2.2 Modeling the Surgical Procedure

In order to simulate a surgical procedure we model the bone cuts (osteotomies) and bone movements (e.g. advancement of the jaw bone) with the help of a craniofacial surgeon using the Alias\textsuperscript{TM} modeling system. Figure 7 depicts the advancements performed on upper and lower jaw bone structures of our example patient.

![Figure 7: Reference points used to model craniofacial surgery with the Alias\textsuperscript{TM} modeling system](image_url)

In order to reposition the upper jaw three reference points are used: the i-point, situated between the tips of the upper incisors, and either the tips of the upper canines or the molars 17 and 27 on the right and left side of the upper jaw respectively. The tooth numbering in figure 7 refers to the standard of the World Dental Federation (FDI). The lower jaw is repositioned with regard to the upper jaw in order to reach a neutral occlusion of two millimeters. In some cases an additional genioplasty is indicated where a part of the lower jaw is cut and moved according to figure 7.

Knowing corresponding vertices of pre- and post-surgical skull surface makes it possible to calculate the displacement field which will be input to the finite element engine as prescribed displacement boundary conditions (see section Section 3).

In a second step, we determine regions which will not be changed during the surgical procedure and therefore can serve as additional zero displacement boundary conditions. This is accomplished by using the 3D paint program StudioPaint\textsuperscript{TM} which allows one to draw directly onto a three-dimensional surface. Figure 8 depicts an example setting of the boundary conditions both on the skull and on the facial surface.

![Figure 8: Boundary conditions:](image_url)

3 FEM Model

This section introduces the finite element system employed for surgery simulation. For reasons of readability we briefly review the basic notions of static elastomechanics which are fundamental for the subsequent discussion of the volumetric FEM approach. Then we elaborate the construction scheme for the hybrid $C^0/C^1$-continuous volume interpolation functions designed for our framework. A section on matrix formulation

$$\mathbf{U}(u, v, w) = \begin{bmatrix} B_u(u, v, w) \\ B_v(u, v, w) \\ B_w(u, v, w) \\ U_u(u, v, w) \\ U_v(u, v, w) \\ U_w(u, v, w) \end{bmatrix}$$

![Figure 9: Three-dimensional face with one example 6-node prism element](image_url)

The soft tissue model we use for facial surgery simulation requires the following idealizations:

- Rather than by explicit application of external body ($f^b$) or surface ($f^s$) loading forces soft tissue deformations are invoked by so-called prescribed skull displacements obtained from the surgical procedure (see figure 8c). Using this approach reduces so-called locking effects [1].
- We restrict our model to the laws of linear elasticity, since the displacements and deformations in most craniofacial operations are small in an FEM sense. In addition, we assume the elasticity as being constant, i.e. independent of the stress.
- Tissue parameters like elasticity and incompressibility do not vary throughout an element.

We think of an elastic body $B$ – the pre-surgical face – in a cartesian $X, Y, Z$ coordinate system as depicted in figure 9. The volume is parametrized in $u$ and $v$ over the surface and in $w$ towards the skull. The computational problem consists of finding the displacement function $U$ which describes the displacement resulting from facial surgery at each point of the volume $B$. In our model this function is a priori known at rigid parts of the face and skull (zero displacement boundary conditions $S_0$) as well as on the parts of the skull that were moved during the surgical procedure (prescribed displacement boundary conditions).

We consider the strain $\mathbf{\varepsilon}$

$$\mathbf{\varepsilon} = \begin{bmatrix} \varepsilon_{XX} & \varepsilon_{XY} & \varepsilon_{XZ} & \gamma_{XY} & \gamma_{XZ} & \gamma_{ZX} \end{bmatrix}^T$$

(2)
which is defined by first order derivatives of the displacement field $\mathbf{U}$. The first three components are known as the volumetric strains

$$
\varepsilon_{XX} = \frac{\partial U_t}{\partial X}, \quad \varepsilon_{YY} = \frac{\partial U_t}{\partial Y}, \quad \varepsilon_{ZZ} = \frac{\partial U_t}{\partial Z} \quad (3)
$$

whereas the second three components are denoted as the deviatoric strains

$$
\gamma_{XY} = \frac{\partial U_t}{\partial X} + \frac{\partial U_t}{\partial Y}, \quad \gamma_{YZ} = \frac{\partial U_t}{\partial Y} + \frac{\partial U_t}{\partial Z}, \quad \gamma_{ZX} = \frac{\partial U_t}{\partial X} + \frac{\partial U_t}{\partial Z} . \quad (4)
$$

The strain caused by the displacement results in a corresponding stress $\tau$

$$
\tau = \begin{bmatrix} \tau_{XX} & \tau_{XY} & \tau_{XZ} \\ \tau_{YX} & \tau_{YY} & \tau_{YZ} \\ \tau_{ZX} & \tau_{ZY} & \tau_{ZZ} \end{bmatrix}^T . \quad (5)
$$

For small strains the fundamental relationship between $\tau$ and $\varepsilon$ is established by the constitutive relation

$$
\tau = \mathbf{C} \varepsilon . \quad (6)
$$

$\mathbf{C}$ depends on the elasticity $E$ of the material and on its incompressibility $\nu$, denoted by Young’s modulus and Poisson ratio, respectively:

$$
\mathbf{C} = \begin{bmatrix} 1 - \nu & \nu & \nu \\ \nu & 1 - \nu & 0 \\ 0 & \nu & 1 - \nu \end{bmatrix} \frac{E}{(1 + \nu)(1 - 2\nu)} . \quad (7)
$$

The Poisson ratio varies between zero for fully compressible and 0.5 for fully incompressible materials. Note that the coefficients of $\mathbf{C}$ increase with $\nu \to 0.5$ and, consequently, $\mathbf{C}$ is not defined for $\nu = 0.5$. For the simulation of incompressibility used to model volume preservation we therefore have to use a different formulation which will be described in the next section.

The solution $\mathbf{U}$ of the problem is the configuration with minimal potential energy which corresponds to the equilibrium of internal elastic energy and the work done by external forces. The absence of such forces allows one to establish the equilibrium condition as a zero virtual work which can be formulated as

$$
\int_{V} \overline{\mathbf{E}}^T \mathbf{C} \overline{\mathbf{E}} \ dV = 0 . \quad (8)
$$

The overbar denotes virtual strains caused by virtual displacements. Further mathematical elaborations on the notion of virtual work are omitted for brevity. A detailed discussion can be found in [1].

### 3.2 Mixed Formulation

In order to deal with incompressible materials the strain has to be separated into its volumetric and deviatoric components and in addition to the displacement function $\mathbf{U}$ the pressure $p$ needs to be introduced as an additional variable. This approach is termed the finite element mixed formulation [1] and will briefly be reviewed in the following.

The constitutive equation (6) can be reformulated using individual notation by separating volumetric and deviatoric components

volumetric part

$$
\tau_{ij} = \kappa \varepsilon_{ij} + 2G \varepsilon_i^e \delta_{ij} , \quad i, j \in \{X, Y, Z\} \quad (9)
$$

deviatoric part

with the new tissue parameters bulk modulus $\kappa$ and shear modulus $G$ which follow from $E$ and $\nu$ to

$$
\kappa = \frac{E}{3(1 - 2\nu)} \quad \text{and} \quad G = \frac{E}{2(1 + \nu)} . \quad (10)
$$

Further in (9), $\delta_{ij}$ is the Kronecker delta, while $\varepsilon_i^e$ and $\varepsilon_V$ denote the deviatoric and volumetric strain, respectively:

$$
\varepsilon_i^e = \varepsilon_{XX} + \varepsilon_{YY} + \varepsilon_{ZZ} = \frac{\Delta V}{V} . \quad (11)
$$

$$
\varepsilon_V = \varepsilon_{XX} + \varepsilon_{YY} + \varepsilon_{ZZ} = \frac{\Delta V}{V} . \quad (12)
$$

The volumetric strain approximates the proportional change of the volume of the body denoted by $\frac{\Delta V}{V}$. Relating a change in volume to a change in pressure we find

$$
p = -\kappa \varepsilon_V , \quad (13)
$$

which can be used to reformulate (9) as

$$
\tau_{ij} = -p \delta_{ij} + 2G \varepsilon_i^e . \quad (14)
$$

Note that for fully incompressible materials $\kappa$ in (13) is infinite but the pressure $p$ is still defined.

Hence, the virtual work of (8) converts to

$$
\int_{V} \overline{\mathbf{E}}^T 2G \varepsilon_i^e \ dV - \frac{1}{V} \int_{V} \varepsilon_V p \ dV = 0 . \quad (15)
$$

In order to establish a relationship between the independent variables pressure and displacement in (15) we introduce (13) written in integral form as a second equation:

$$
\int_{V} \left( \frac{\mathbf{B}^T \varepsilon_i^e}{\kappa} + \varepsilon_V \right) p \ dV = 0 . \quad (16)
$$

We can think of $\mathbf{p}^\mathbf{B}$ in (16) as a Lagrange multiplier enforcing the constraint (13) between pressure and displacement.

If both equation (15) and (16) are fulfilled the displacement field $\mathbf{U}$ provides a solution to the problem and the resulting post-surgical face is given by $\mathbf{B}^\mathbf{U} = \mathbf{B} + \mathbf{U}$.

In the following section we derive a set of interpolation functions which will be used to expand the solution $\mathbf{U}$ within a finite element in a Galerkin type approach to discretize (15) and (16).

### 3.3 Interpolation Functions

The design of the volumetric interpolation functions used in our approach was motivated both by the demand to conform to the underlying physics and by the need for visually appealing facial surfaces. In order to satisfy both aims we come up with a set of prismatic interpolation functions featuring $C^1$-continuity at the facial surface. Both the interior and the boundary triangle at the
bottom – representing a patch of the skull surface – are left to be \( C^0 \)-continuous in order to reduce the overall number of degrees of freedom (DOF) of the system. The prismatic topology of the elements specifically simplifies the FEM mesh generation. The element and all DOFs are depicted in figure 10a.

In order to achieve \( C^1 \)-continuity at the surface we first replace \( P_1 \) by a set of nine functions derived from the well-known \( \mathbf{N}^9 \) introduced by \cite{5,15}. Thus we arrive at a set of functions \( \mathbf{P}^{12a} \) which allows to control the derivatives around the vertices on the surface. The resulting Hermite type barycentric polynomial functions for vertex \( R \)

\[
\begin{align*}
P^{12a}_1 &= (r + r^2 s + r^2 t - r^2)(1 - q) \\
P^{12a}_2 &= (r^2 s + 1 r t)(1 - q) \\
P^{12a}_3 &= (r^2 + 1 r t)(1 - q)
\end{align*}
\]

are depicted in figure 11. The remaining functions \( P^{12a}_4 \ldots P^{12a}_9 \) can be obtained by a cyclic permutation of \( r, s, t \).

\[
(18)
\]

\[
(19)
\]

\[
(20)
\]

\[
(21)
\]
where \( u_i \) and \( v_i \) denote the coordinates of the vertices of the surface triangle. As an example, figure 12 shows the resulting \( P_2^{12b} \) interpolation function defined over a one-ring triangulation whose center vertex has valence six.

The contour line representation in figure 12b illustrates that although providing \( C^1 \)-continuity at vertices the \( P_2^{12b} \) still suffer from discontinuities between adjacent elements. In finite element analysis, rational blend functions are used to tackle this problem (see e.g. [23]). The following rational \( e \) functions allow us to control the cross-boundary derivatives independently of the displacements and derivatives at the vertices.

They are initially defined for the trivariate barycentric setting as follows:

\[
e_1 = \frac{r^2 s^2 t^2 (1 + r)}{(r + s)(r + t)} \\
e_2 = \frac{r^2 s^2 t^2 (1 + s)}{(s + t)(s + r)} \\
e_3 = \frac{r^2 s^2 t^2 (1 + t)}{(t + r)(t + s)}.
\]  

(22)

Figure 13a depicts a linear volumetric tensor product extension of \( e^1 \) by a multiplication with \((1 - q)\). The resulting function controls the cross-boundary derivative at the edge \( r = 0 \).

A straightforward method to develop \( C^1 \)-continuous surfaces is to integrate the required weights of the \( e_i \) functions into the FEM problem as additional variables. Although being popular [5,15], a major disadvantage of this approach is the increase of the overall degrees of freedom of the global system leading to higher computational costs. Further, various special cases have to be dealt with during the finite element assembly.

Therefore, we developed the following \( N_9^s \) interpolation functions for the prism surface according to [23]. Instead of introducing the weights of the \( e \) functions as DOFs they are computed from the cross-boundary derivatives of the \( N^9 \) functions at the vertex positions. We start with the normal derivative across each edge defined as

\[
\frac{\partial}{\partial n} e_1(0, 0.5, 0.5) = \frac{l_1}{8\Delta} \\
\frac{\partial}{\partial n} e_2(0.5, 0, 0.5) = \frac{l_2}{8\Delta} \\
\frac{\partial}{\partial n} e_3(0.5, 0.5, 0) = \frac{l_3}{8\Delta}.
\]

(23)

As a next step we normalize the slopes of the \( e_i \) to yield the \( e'_i \) prism functions with unit cross-boundary derivatives

\[
\frac{\partial}{\partial n} e'_1(0, 0.5, 0.5) = \frac{8\Delta e_1}{l_1} \\
\frac{\partial}{\partial n} e'_2(0.5, 0, 0.5) = \frac{8\Delta e_2}{l_2} \\
\frac{\partial}{\partial n} e'_3(0.5, 0.5, 0) = \frac{8\Delta e_3}{l_3}.
\]

(25)
In order to obtain the $C^1$-continuous $P^{12}$ interpolation functions we compute the average of the corresponding cross-boundary derivative $\frac{\partial}{\partial n_i}$ at the endpoints of each edge $i$ of the surface triangle for the first nine components of $P^{12b}$:

\[
Y = \frac{1}{2} \begin{bmatrix}
\frac{\partial}{\partial n_1} P^{12b}(1, 0, 0, 0) + \frac{\partial}{\partial n_2} P^{12b}(1, 0, 0, 0) + \frac{\partial}{\partial n_3} P^{12b}(1, 0, 0, 0) \\
\frac{\partial}{\partial n_2} P^{12b}(0, 1, 0, 0) + \frac{\partial}{\partial n_3} P^{12b}(0, 1, 0, 0) \\
\frac{\partial}{\partial n_3} P^{12b}(0, 0, 1, 0) + \frac{\partial}{\partial n_1} P^{12b}(0, 0, 1, 0)
\end{bmatrix}.
\] (26)

After computing the cross-boundary derivative at the edge midpoints of the first nine components of $P^{12b}$

\[
Z = \begin{bmatrix}
\frac{\partial}{\partial n_1} P^{12b}(0, 0, 1, 0) \\
\frac{\partial}{\partial n_2} P^{12b}(0, 0, 1, 0) \\
\frac{\partial}{\partial n_3} P^{12b}(0, 0, 1, 0)
\end{bmatrix}
\] (27)

we are now able to define a vector of prism shaped interpolation functions featuring global $C^1$-continuity at the top surface

\[P^{12} = P^{12b} + e(Y - Z) | 0].
\] (28)

Figure 13b depicts the contour lines for the one-ring example of figure 12a using the $P^{12}$ shape functions for interpolation. All discontinuities are resolved.

**Element Node Vector**

In contrast to the interpolation of the vector valued displacement with the $P^{12}$ functions the $P^6$ linear functions are used to interpolate the scalar variable representing the pressure in the mixed formulation.

The upper definitions yield the following node or weight vector for an element $m$

\[
[\hat{U} | P]^{(m)} = \left[ U_x^{(m)}, U_y^{(m)}, U_z^{(m)}, P^{(m)} \right]
\] (29)

with the weight subvectors $\hat{U}_x^{(m)}$, $\hat{U}_y^{(m)}$ and $\hat{U}_z^{(m)}$ for the interpolation of displacements in the corresponding coordinate directions and the weights $P^{(m)}$ for the interpolation of pressure:

\[
\begin{align*}
\hat{U}_x^{(m)} &= \begin{bmatrix} \frac{\partial U_x^{(m)}}{\partial u_1} & \frac{\partial U_x^{(m)}}{\partial u_2} & \frac{\partial U_x^{(m)}}{\partial u_3} & \frac{\partial U_x^{(m)}}{\partial u_4} & \frac{\partial U_x^{(m)}}{\partial u_5} & \frac{\partial U_x^{(m)}}{\partial u_6} \end{bmatrix} \\
\hat{U}_y^{(m)} &= \begin{bmatrix} \frac{\partial U_y^{(m)}}{\partial u_1} & \frac{\partial U_y^{(m)}}{\partial u_2} & \frac{\partial U_y^{(m)}}{\partial u_3} & \frac{\partial U_y^{(m)}}{\partial u_4} & \frac{\partial U_y^{(m)}}{\partial u_5} & \frac{\partial U_y^{(m)}}{\partial u_6} \end{bmatrix} \\
\hat{U}_z^{(m)} &= \begin{bmatrix} \frac{\partial U_z^{(m)}}{\partial u_1} & \frac{\partial U_z^{(m)}}{\partial u_2} & \frac{\partial U_z^{(m)}}{\partial u_3} & \frac{\partial U_z^{(m)}}{\partial u_4} & \frac{\partial U_z^{(m)}}{\partial u_5} & \frac{\partial U_z^{(m)}}{\partial u_6} \end{bmatrix} \\
P^{(m)} &= \begin{bmatrix} \frac{\partial P^{(m)}}{\partial u_1} & \frac{\partial P^{(m)}}{\partial u_2} & \frac{\partial P^{(m)}}{\partial u_3} & \frac{\partial P^{(m)}}{\partial u_4} & \frac{\partial P^{(m)}}{\partial u_5} & \frac{\partial P^{(m)}}{\partial u_6} \end{bmatrix}
\end{align*}
\] (30)

**Initial surface fairing**

As we are computing a smooth displacement field to the original configuration we require a smooth presurgical facial shape. Therefore, we have to find estimates for an initial node vector used to define the pre-surgical smooth shape.

In the proposed setting, it would be straightforward to use the finite element approach to compute a $C^1$-surface with minimal energy with respect to a surface energy measure $[5, 15]$. This, however, would require an additional FEM solving step in advance. In order to reduce the computational costs we approximate the nodal weights corresponding to the derivatives of the initial surface by means of finite differences. Alternatively, subdivision schemes such as [13] and discrete fairing methods like [11] could be used to find estimates.

### 3.4 Matrix Formulation

Using the sets of shape functions $P^{12}$ and $P^6$ together with the corresponding weight vectors (30) yields

\[
\begin{bmatrix} U^{(m)}(u, v, q) \\
U_x^{(m)}(u, v, q) \\
U_y^{(m)}(u, v, q) \\
U_z^{(m)}(u, v, q)
\end{bmatrix} = \begin{bmatrix} P^{12(m)} \\
0 \\
0 \\
0
\end{bmatrix} U^{(m)} = H^{(m)} U
\] (31)

for the interpolation of displacements and

\[
P^{(m)(u, v, q)} = P^6 P^{(m)} = H P^{(m)}
\] (32)

for the interpolation of pressure within an element $m$. Note that in contrast to the $P^{12}$ the linear shape functions $P^6$ are independent of the element geometry and therefore can be written as $H_p$ instead of $H^{(m)}$.

By introducing an operator matrix of the first order derivatives used in the definition of the strain vector the deviatoric strain (11) can be formulated as

\[
\varepsilon = \frac{1}{3} \begin{bmatrix} 2 & -1 & -1 & 0 \\
-1 & 2 & -1 & 0 \\
-1 & -1 & 2 & 0 \\
0 & 0 & 0 & 3
\end{bmatrix} \begin{bmatrix} \frac{\partial \hat{U}}{\partial x} \\
\frac{\partial \hat{U}}{\partial y} \\
\frac{\partial \hat{U}}{\partial z} \\
\frac{\partial P^{(m)}}{\partial x}
\end{bmatrix}^T
\] (33)

and for the volumetric strain (12) we find analogously

\[
\varepsilon = \begin{bmatrix} 1 & 1 & 1 & 0 \\
1 & 1 & 1 & 0 \\
1 & 1 & 1 & 0 \\
0 & 0 & 0 & 3
\end{bmatrix} \begin{bmatrix} \frac{\partial \hat{U}}{\partial x} \\
\frac{\partial \hat{U}}{\partial y} \\
\frac{\partial \hat{U}}{\partial z} \\
\frac{\partial P^{(m)}}{\partial x}
\end{bmatrix}^T
\] (34)

Using the Galerkin projections (32), (33) and (34) to express (15) and (16) yields a matrix formulation of the problem integrating over each element using Gaussian quadrature and summing up the contribution of each element into one global system of linear equations yields

\[
K U = 0
\] (35)

where $K$ denotes the global stiffness matrix and $U$ represents the global weight vector we are solving for. Invoking the prescribed displacement boundary conditions yields a non-trivial solution of (35). Either a conjugate gradient approach or a direct solver for the case of total or near incompressibility is used to solve.
(35). For a survey of matrix computations see e.g. [9]. An elaboration of the structure and derivation of (35) is given in the appendix.

4 RESULTS AND VALIDATION

Error Evaluation and Visualization

In order to validate the prototype and to obtain both a quantitative and a visual impression of the quality of simulation, we quantitatively compare the post-surgical facial surface with the surface resulting from simulation. Therefore, we first register both geometries as described in section Section 2.1 and then approximate the error at each surface coordinate by projecting the radial distance between the surfaces onto a local normal at the intersection points. The error visualization as depicted in figure 14a is obtained by pseudo-coloring these distance values at the facial areas affected by surgery. Non-matchable regions like hair, eyebrows and eyes are set to zero.

Figure 14: Visualization of the error distribution in terms of the radial distance between simulated and real post-surgical surface (a) and the color map used for pseudo-coloring (b)

The pseudo-coloring is performed by means of the color map in figure 14b. Positive values indicate the predicted surface lying in front of the post-surgical face whereas red colors illustrate the contrary. Errors at the side of the nose common to all cases must be attributed to the radial measuring set-up which tends to overemphasize errors at places where the surface happens to be of nearly the same orientation as the radial direction of measurement. Further, errors at the cheekbones are due to swelling. The quantitative simulation error given as a volumetric difference in figures Section 15 to Section 18 is computed by integrating the absolute values of the distances over the surface of interest.

Sources of Error

According to clinicians, the prediction performance of our system is excellent. However, before presenting individual cases some general sources of errors deserve discussion. Detectable minor differences between prediction and real result may be contributed to the following factors:

- The exact correspondence of the three-dimensional metric displacements such as depicted in figure 7 with the actual surgery is of paramount importance for the quality of simulation.
- Slight deviations may result from the state of swelling which can take up to one year. Further, differences in facial muscular activity between pre- and post-operative picture and laser range scan are unavoidable.
- The matching of the post-surgical laser range scan and the result of the simulation needed for the quantitative evaluation is prone to errors as appropriate reference points can only be found in unaltered regions of the face. In addition, head alignment and motion artifacts can distort the quality of the laser range scans.
- Overemphasis of certain facial structures in the resulting renderings of the simulation is due to differences in the lighting conditions of simulation and post-surgical photographs.

Individual Patients

The results of the simulated correction of the short face syndrome of our example patient are depicted in figure 15. He suffers from a retrodisplacement of the maxilla and mandible (upper and lower jaw) in combination with a deep bite because of a predominantly horizontal growth pattern of the bases of the jaws in relation to the skull base.

The correspondence of simulation and real surgery is exceptional. The profile lines are given in figure 15d, where blue, green and red represent pre-surgical, simulated and post-surgical situation respectively. Minor deviations can only be observed in the region of the mouth. In addition, the frontal error visualization reveals artifacts due to swelling.

In the second case (figure 16) a retromaxillism with dished-in midface had to be cured. This malformation is caused by an insufficient anterior (sagittal) growth of the maxilla in combination with a slightly overprojected mandible. As a consequence, upper lip, nose and especially paranasal tissues are not sufficiently advanced. Further, the chin is far to prominent.

Again, there is little deviation between simulation and real outcome. The main error at the lower chin is caused by unaligned scanning positions.

Figure 17 shows a complex facial asymmetry due to a chronic disease of the facial bones in the left midface (fibrous dysplasia). Overgrowth of the left half of the lower jaw, the left cheekbone and the bones surrounding the eye sockets (periorbital bones) led to a severe transversal deviation of the occlusal plane.

The simulation of a case of such complexity is demanding due to the correction of asymmetries as well as because of the rotations and the genioplasty introduced during surgery. Although, as can be seen in the profile lines, the correspondence at the chin and lips is satisfactory.

Similar to the case in figure 16, figure 18 shows a patient suffering from excessive vertical growth of the retrodisplaced maxilla. In this case, only the molar teeth are in contact. The incisor teeth are too far behind and therefore cannot touch their antagonists in the mandible. In addition, the lower jaw is overprojected to a certain extent.

The surgical procedure only introduced a very small displacement field. Although the resulting contour lines are acceptable the accurate simulation of such small bone movements lies beyond the resolution bounds of the system.

Computational Costs

It has to be stated, that the actual computation times are substantially smaller compared to the model build-up and the modeling of the surgical procedure which can take up to four hours. The FEM computation itself takes about half an hour on an SGI R10000 195 MHz Indigo2 for the model resolution shown in the
results. The resulting sparse system of linear equations is of the approximate size of 100k x 100k with a density of about 10^{-2} to 10^{-1} percent.

5 CONCLUSIONS AND FUTURE WORK

We have presented a framework for facial surgery simulation which combines the quality of a $C^1$-continuous facial surface with the accuracy of volumetric finite element simulation. Furthermore, in a test series with real patient data available in a clinical environment we have given a proof of concept by comparing simulation and real surgery. Both the error analysis and the clinician’s evaluation of the results demonstrate the superiority of our vision for surgical planning over the conventional methods.

In spite of the performance of the framework future work will comprise the following aspects: the restriction to $C^0$-continuity in the interior might have effects on the volumetric behavior of...
the tissue. An extension of the system to $C^1$-continuous tetrahedral elements giving additional topological freedom is currently being investigated. Furthermore, in order to comply with results of biomechanical studies, we plan to include non-linear elasticity into the model. Since, however, the additional computational costs are enormous the resulting benefits have to be evaluated thoroughly. Finally, adaptive multigrid methods in combination with hierarchical bases could substantially speed-up the simulation.

Figure 17: Profiles: (a) pre-surgical, (b) predicted, (c) post-surgical, (d) profile lines
Portraits: (e) pre-surgical, (f) predicted, (g) post-surgical, (h) error visualization

Figure 18: Profiles: (a) pre-surgical, (b) predicted, (c) post-surgical, (d) profile lines
Portraits: (e) pre-surgical, (f) predicted, (g) post-surgical, (h) error visualization

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APPENDIX – Matrix Formulation

In a first step, we substitute (32), (33), and (34) for one element m into (15) and (16). With \( C^{(m)} \) representing the deviatoric stress-strain material matrix following from (9) to

\[
C^{(m)} = 2G^{(m)}I \quad I: 6x6 \text{ identity matrix} \quad (A.1)
\]

this yields for (15)

\[
\bar{U}^{(m)T} \left[ \int_{\omega (m)} B_D^{(m)T} C^{(m)} B_D^{(m)} dV^{(m)} \right] \bar{U} + \\
+ \bar{U}^{(m)T} \left[ - \int_{\omega (m)} B_V^{(m)T} \bar{H}_p dV^{(m)} \right] \bar{P} = 0 \quad (A.2)
\]

and, after multiplication of (16) with -1,

\[
\bar{P}^{(m)T} \left[ - \int_{\omega (m)} \bar{H}_p^{T} H_p^{T} dV^{(m)} \right] \bar{U} + \\
+ \bar{P}^{(m)T} \left[ - \int_{\omega (m)} \bar{H}_p^{T} B_V^{(m)} dV^{(m)} \right] \bar{P} = 0 \quad (A.3)
\]

Subsequent repeated application of the principle of virtual displacement yields the following system of linear equations the solution of which fulfills the conditions (A.2) and (A.3) within an element m

\[
\begin{bmatrix}
K_{uu}^{(m)} & K_{up}^{(m)} \\
K_{pu}^{(m)} & K_{pp}^{(m)}
\end{bmatrix}
\begin{bmatrix}
\bar{U}^{(m)} \\
\bar{P}^{(m)}
\end{bmatrix} = 0 \quad (A.4)
\]

with the stiffness submatrices corresponding to (A.2) and (A.3)

\[
K_{uu}^{(m)} = \int_{\omega (m)} B_D^{(m)T} C^{(m)} B_D^{(m)} dV^{(m)} \\
K_{up}^{(m)} = \int_{\omega (m)} B_D^{(m)T} \bar{H}_p dV^{(m)} \\
K_{pu}^{(m)} = \int_{\omega (m)} B_V^{(m)T} \bar{H}_p dV^{(m)} \\
K_{pp}^{(m)} = \int_{\omega (m)} H_p^{T} H_p^{T} dV^{(m)} \quad (A.5)
\]

Evaluating the integrals in (A.5) using Gaussian quadrature and summing up the contributions of all elements – a process known as the finite element assembly – finally results in the global system of linear equations (35)

\[
KU = \begin{bmatrix}
K_{uu} & K_{up} \\
K_{pu} & K_{pp}
\end{bmatrix}
\begin{bmatrix}
\bar{U} \\
\bar{P}
\end{bmatrix} = 0 \quad (A.6)
\]

where the submatrices and subvectors are the global counterparts to the local matrices and vectors in (A.4).

REFERENCES