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A REVIEW OF FINITE ELEMENT APPLICATIONS IN ORAL AND MAXILLOFACIAL BIOMECHANICS

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Finite element method (FEM) is preferred to carry out mechanical analyses for many complex biomechanical structures. For most of the biomechanical models such as oral and maxillofacial structures or patient-specific dental instruments, including nonlinearities, complicated geometries, complex material properties, or loading/boundary conditions, it is not possible to accomplish an analytical solution. The FEM is the most widely used numerical approach for such cases and found a wide range of application fields for investigating the biomechanical characteristics of oral and maxillofacial structures that are exposed to external forces or torques. The numerical results such as stress or strain distributions obtained from finite element analysis (FEA) enable dental researchers to evaluate the bone tissues subjected to the implant or prosthesis fixation from the viewpoint of (i) mechanical strength, (ii) material properties, (iii) geometry and dimensions, (iv) structural properties, (v) loading or boundary conditions, and (vi) quantity of implants or prostheses. This review paper evaluates the process of the FEA of the oral and maxillofacial structures step by step as followings: (i) a general perspective on the techniques for creating oral and maxillofacial models, (ii) definitions of material properties assigned for oral and maxillofacial tissues and related dental materials,

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(iii) definitions of contact types between tissue and dental instruments, (iv) details on loading and boundary conditions, and (v) meshing process.

Keywords: Finite element modeling; finite element analysis; oral and maxillofacial biomechanics; mechanical analysis; numerical solution.

1. Introduction

Although the dentistry is a branch of medical sciences, it has a direct relationship with the mechanics of materials. Dental treatments such as dental implantation due to a tooth loss, fixation of fractured bony structures, prosthodontic rehabilitations or even orthodontic solutions require biomechanical perspective. All materials used by dentists for the treatment of such problems should be tested from the biomechanical point of view. Moreover, in addition to the appropriate biomechanical behaviors, such materials should also be biocompatible, thereby leading to restriction of the existing materials that may meet the required mechanical criteria for dental applications. There are many different methods described for the evaluation of the biomechanical behaviors of different materials during the development of modern dentistry. In recent years, a dramatic improvement has been observed among these biomechanical analysis methods, in parallel with the development of the technology.

There is a variety of existing mechanical testing methods to examine the mechanical properties of the structures such as photoelasticity, strain gauge-based measurements, optic measurement, and computational approaches.[1,2] Probably the most common method is the strain gauge which is based on the measurement of the alteration of electrical resistivity. It is easy to install and provides precise strain results under different operating conditions. However, strain gauges provide data only at the points where they are mounted and therefore it is not possible to obtain a distributed strain data over the surface of the object. Stress field over a whole surface can be experimentally obtained by photoelasticity. On the other hand, expensive equipment is a need for the precision analysis of large components using this technique.[3] Numerical techniques are more flexible for analyzing the stress or strain alterations than the experimental ones. Moreover, computational methods are easier and more flexible to compare the effects of some parameters regarding the material properties of the objects of interest.

Finite element method (FEM) is the most widely preffered computational technique used in the biomechanics for the mechanical analysis of the hard tissue models. This method has become very popular especially over the last decade, thanks to the impressive technological developments associated with the programs enabling computer-aided design and creating three-dimensional models from imaging data. Today, finite element analysis (FEA) is an integral part of many design and manufacturing processes, especially in the various areas of dental biomechanics. Some of the widely investigated contents within the dental biomechanics are (i) the stress and strain distribution around the interconnected surfaces of bone tissues and dental implants,[4–10] (ii) mechanical evaluation of strengths of the materials used for dental instruments,[1,11–15] and (iii) finding optimum location of implants and miniplates and reliable implants and miniplates type for treatment to obtain the optimum stress-strain distributions over bone tissue and the material.[16–24]

Since the oral and maxillofacial models are of nonlinear characteristics, a large variety of biomechanical properties and complicated geometries, most of the time, an analytical solution cannot be achieved. FEM enables researchers an easier way to estimate the biomechanical behavior of the tissues or instruments of interest than in vivo measurements. It provides stress, strain, energy, displacement and pressure data at any location. The distribution of such data on models can be visualized by specific software.[25–33]

FEM can solve unknowns of the model with complex constitutive laws of material behaviors (e.g., articular disk, ligament etc.), and various load and boundary conditions (e.g., external force, thermal change and magnetic field power by a series of computational procedures). In this process, the model in the global domain is divided into small and finite domains (elements) with differential equations.[34] The iterative incremental solution is done by solving the problem. If the model is not well-conditioned, the convergence will be slower.

Mechanical analysis of biomechanical models with nonlinear load, material, geometry and contact properties can be carried out by FEA. Loading nonlinearity describes time-dependent loading that is causing large displacement. Material nonlinearity comes from plasticity, viscoplasticity, and creep properties of materials. Nonlinear contact properties are associated with frictional slip surface.

2. Finite Element Analysis of Oral and Maxillofacial Structures

The first step of the FEA process is to obtain a 3D model of the object which is intended to be analyzed. The accuracy of analysis can be improved by constructing detailed geometry of the model, defining realistic material properties and load/boundary conditions, establishing true contact and joint types, as well as providing a well-structured mesh. The accuracy of the model specifies the reliability of the results obtained from FEA. If these properties can be defined in a realistic fashion without important assumptions, the solution of the analysis will be close to those seen in the actual case.

2.1. *Modeling*

Conversion of cross-sectional computed tomography (CT) images into 3D solid body models is the most widely used model generation method for biologic structures.[35–38] Different biologic tissues absorb radiation at different levels which causes different effects on tomography data related with the radio-density of the tissue. These radio density differences among anatomic structures are processed by tomography software by using 'Hounsfield Scale' which is a quantitative scale for describing radio density.[39–41] By means of hounsfield unit (HU) scale, different images of different anatomic structures can be defined on the computed tomography data. In this way, different anatomic structures having different HU values can be converted into 3D volumetric models separately and consequently, more detailed anatomic computer aided design (CAD) models having specific material properties can be generated. By defining threshold values, most of the tissues forming the human body such as cortical or trabecular bone tissue, enamel, dentin, pulp, and suture can be separated from each other. Threshold values ranged between 280 and 3071 HU can be defined to separate the mandible from the soft tissues.[8] Briefly, a realistic geometry and material assignment can be done by CT scan data.

The other method for modeling is that using micro CT (μCT) data, especially for the geometries with micro dimensional details.[11] It is also possible to extract the solid body models of biological structures from magnetic resonance imaging (MRI) data that be directly processed by 3D reconstruction software. It is especially useful to separate soft tissues from whole scanned tissues of the body. One drawback of MRI is that images should be on the same plane; the other one is that the extraction of the indented tissue from other tissues is quite a time-consuming process.[28,42,43]

Images obtained from digital scanners can also be used for modeling in FEA with a high resolution of visualization. However, only synthetic bone models or dry bones from cadavers can be scanned by means of these devices, thereby providing only outer surfaces of the model that it cannot characterize many mechanical properties of the real biological tissues.[44]

The skeleton system is made of two types of osseous tissues i.e., compact (cortical) and cancellous bones, which differ in their structure, function, and distribution. The compact bone covers the mandible as a shell. The almost entire volume inside that shell is filled by cancellous bone. The cancellous bone is responsible for vital functions and compared to the compact bone from the mechanics of materials point of view, it has a higher surface area but is less stiff and dense, and weaker.[5,8,45] On the other hand, the cortical layer, in contrary to the cancellous part, has limited vital functions besides its high mechanical durability. In addition, cancellous tissue has a more complex anatomic shape by having Haversian canals, Wolkman canals, and canaliculus than the cortical layer, which makes the 3D modeling process of the cancellous tissue more complicated and, thereby causing failures in FEA. Furthermore, it was shown that the cancellous layer does not contribute significant rigidity and all the strength of the bone can be attributed to the cortical layer.[46] Therefore, there is a general tendency to take only the compact layer of the mandible into account for FEA models by neglecting the biomechanical behaviors of cancellous bone.[47,48]

In many biomechanical studies regarding oral and maxillofacial structures, mandible and maxilla are among the most widely analyzed parts in terms of the mechanical properties. The mandible has more compact (cortical) bone tissue than the maxilla due to having exposed high level of muscle forces in amplitude and its moveable nature. On the other hand, the maxilla is different from the mandible in terms of including different amounts and density of cortical and cancellous layers. Although the cortical parts of the zygomatic alveolar buttress and nasoalveolar processes are thicker than the other cortical parts of the maxilla, the thickness of the cortical layer of the maxilla is relatively small. In addition, generally, cancellous part of the maxilla is denser than the mandibular cancellous tissue and it is biomechanically stronger. As a result, accurately modeling the cancellous and cortical layers of the maxilla and mandible are crucial to evaluate the biomechanical responses of these tissues when facing with external loadings. However, it is very complicated and prone to failures due to complex geometry and heterogeneous distribution of the spongious bone.

Thickness of the compact bone tissue alters heterogeneously across mandible. In literature, cortical layer within mandible was modeled as uniformly with a thickness of 1,[1,4,47,49–51] 0.5,[15,52] 1.5,[20,53] 2,[22,48,54,55] and 5 mm.[56] As for the maxilla, the cortical layer was generated as uniformly distributed with thickness of 1[19,57] and 2 mm.[1]

Periodontal ligament (PDL) is a connective tissue and connects the teeth and alveolus. Functional role of this tissue is generally accepted as tooth support. In FEA studies regarding dental biomechanics, properties of the PDL model have an important influence on the numerical results. In many studies, the PDL was modeled as a thin homogeneous layer surrounding the tooth with thickness of 0.2 mm,[15,16,58] and 0.25 mm.[7,11]

Some researchers prefer to study on small models to simplify the shapes or the materials of the corresponding biological structure. For example, Ref. 35 extracted the central incisors and surrounding bone from mandible for simplification. On the other hand, in such cases, stress distribution would be in limited visualization and this would restrict the obtained results from being an accurate representation of the actual phenomenon. However, complex shapes of the bone model would increase the duration of the modeling process and the computational cost.

2.2. *Materials*

Material assignment in FEA is an important phase of the analysis. Isotropic materials have the mechanical and thermal properties that are the same in all directions. In contrast, the anisotropic materials show different mechanical and thermal characteristics along different directions under the same loading conditions. The orthotropic materials, which are a subgroup of the anisotropic materials, have the different material properties along three mutually-orthogonal axes.

2.2.1. *Isotropic materials*

In FEA software, specifying the elastic modulus (Young's modulus) and Poisson's ratio would be enough for identification of the material. The elastic modulus of bones can be predicted from CT data. The material properties can be characterized by HU values. These coefficients are related to tissue density. Density values can be converted to elastic modulus values by means of different approaches by relating the

HU value, elastic modulus and apparent bone density to each other.[46] HU values of the bone can be measured using certain software and the Young's modulus can be estimated by some calculations.[59] The relation between HU and apparent density can be established by Eq. (1), where density ρ is expressed in g/cm^3.[39]

$$\rho = 0.007764\text{HU} - 0.056148. \tag{1}$$

Morgan *et al.* (2003) proposed Eq. (2) in order to relate between density ρ expressed in g/cm^3 and elastic modulus E expressed in GPa for femoral neck.

$$E = 6.850 \times \rho^{1.49}. \tag{2}$$

In a study of Wirtz *et al.* (2000), averaged elasticity-density relationship for cancellous bone calculated by Eq. (3) and for cortical bone by Eq. (4) were given from the data in the literature (specimens loaded in the axial direction).[60]

$$E_{\text{cancellous}} = 1.904 \times \rho^{1.64}, \tag{3}$$

$$E_{\text{cortical}} = 2.065 \times \rho^{3.09}. \tag{4}$$

The method of estimation of the elastic modulus from HU values enables researchers to assign specific mechanical properties to bone sections. Therefore, FE models can be constructed more detailed than the models that only constitutes by cortical and cancellous layers each having constant elastic modulus. Table 1 represents an extended summary of FEA studies. For each cited study, respective elastic modulus and Poisson's ratio (ν) values assigned to various biological tissues are reported.

PDL acts like a spring element. The PDL cannot be defined easily in FE models because of its viscoelastic, nonlinear, anisotropic, and nonhomogeneous behavior. Material properties of the PDL were obtained with difficulty in the experimental studies. Ruse (2008) reported that most studies used incorrect elasticity modulus of PDL and stated that "*Although Yettram wrote* 6.89×10^{-5} *GPa in* 1977*, the three order of magnitude error in Weinstein's paper wrote as* 6.89×10^{-2} *GPa in* 1980".[70]

As for the materials used for dental instruments, Table 2 summarizes the values of the elastic modulus and Poisson's ratio that are described in the literature of FEA studies.

Due to its appropriate properties such as inertness, biocompatibility, ductility, and malleability to endodontics, gutta percha (GP) is commonly used as filling material for filling holes in endodontic applications.[35] Ruse (2008) reported the same issue regarding the PDL for GP, namely it has been described with erroneous elasticity modulus in most studies.[35,61] With regards to GP, Ruse (2008) stated that the value of 6.9×10^{-4} GPa was incorrect and the correct values ranged two to three orders of magnitude higher.[70]

The commonly used Poisson's ratio and elastic modulus values in FEA are given in Table 3 for auxiliary materials. Despite their negligible effect in comparison with bone and dental instruments, mechanical properties of the auxiliary materials should also be carefully described.

Table 1. Elastic modulus and Poisson's ratio of isotropic tissues.

Material	Elastic modulus-E (MPa)	Poisson's ratio (ν)
Cortical bone	13000[9,30,49]	0.26[64]
	13700[1,10,12,14,22,25,28,42]	0.3[1,9,10,12,13,22,25,28,30,42,49,52,61–63]
	14500[29]	0.323[29]
	15000[46]	0.33[46]
	16700[13]	
	17800[63]	
	20700[52]	
	34000[64]	
Cancellous bone	345[30,65]	0.225[44]
	480[44]	0.3[10,12,13,25,29,30,42,49,52,61,62]
	759[13]	0.31[65]
	1000[42]	
	1370[10,12,22,25,29,57,61,62]	
	1600[49]	
	14800[52]	
Enamel	72700[66]	0.20[65]
	80000[11,26]	0.30[7,11,26,28]
	84100[7,25,28,65]	0.33[66]
Dentin	18000[61,67]	0.31[7,11,25,26,28,61,65–67]
	18600[7,11,25,26,28,65,66]	
Pulp	2[12,26,28,66,68]	0.45[7,12,26,28,66,68]
	6.8[7]	
Periodental ligament	0.0689[11,66]	0.45[12,25,28,61,64,65,69–71]
	0.05[18]	0.49[16,18,72]
	0.3[71]	
	0.5[61]	
	0.68[7]	
	1[70]	
	27[69]	
	50[16,65,72]	
	68.9[12,25,58]	
Gingiva	0.019[61]	0.3[2,61,64]
	19.6[2,64]	
Alveolar bone	2000[18]	0.3[18,28]
	11500[28]	
Suture	68.65[16]	0.40[16]
Articular disc	6[75]	0.4[74]
	44.1 (Stress $<$ 1.50 MPa)[73]	
	92.4 (Stress $>$ 1.50 MPa)[73]	
	100[74]	
Sinus membrane	58[62]	0.45[62]
Bone	2000[15]	0.3[15,24]
	14800[24]	
Bone graft	11000[62]	0.3[62]
Chondroid ossification	10000[62]	0.3[62]
Low cancellous	800[57]	0.3[57]
Muscle	1[33]	0.3[33]
Mucosa	0.01[34]	0.4[34]
	0.021[76]	0.37[76]
Bone out of the interforaminal region	4500 [76]	0.3[76]

Table 1. (*Continued*)

Material	Elastic modulus-E (MPa)	Poisson's ratio (ν)
Initial blood and granulation tissue	1[62]	0.17[62]
Between initial blood and soft callus	500[62]	0.3[62]
Soft callus	1000[62]	0.3[62]
Between soft callus and stiff callus	3500[62]	0.3[62]
Stiff callus	6000[62]	0.3[62]

Table 2. Elastic modulus and Poisson's ratio of dental instruments.

Material	Elastic modulus-E (MPa)	Poisson's ratio (ν)
Gold	89500[11,26,61]	0.3[53]
	100000[53]	0.33[11,26,61]
Zirconium	205000[11]	0.22[11]
	210000[1]	0.33[1]
Titanium	104000[13,44,61]	0.3[13]
	107000[52]	0.33[20,61]
	110000[2,20,30,35]	0.34[1,42,44,46,52,67]
	113800[67]	0.35[30,35]
	114000[1,42]	
	115000[46]	
Stainless steel	193000[67]	0.3[15,16,18,67]
	200000[15,18,61]	0.33[61]
	205900[16]	
Chromium–cobalt alloy	218000[11]	0.31[11]
	220000[2]	0.3[2]
Feldspathic ceramic	69000[26]	0.19[11]
	70000[11]	0.30[26]
Ceramic, Ips empress	65000[66]	0.19[66]
	68900[50]	0.28[50]
Ni–cr	205000[26,67]	0.33[26,67]
Zinc phosphate	13700[67]	0.33[67]
	22400[26]	0.35[26]
Porcelain	67700[10]	0.22[64]
	68900[35,53]	0.28[10,35,53]
	70000[64]	0.33 2[9]
	82800[1,29]	0.35[1]
Resorbable material [copolymer of l-lactide (17%), d-lactide (78.5%), and tmc monomers (4.5%)]	3150[21,46]	0.46[21,46]
Feldsphatic porcelain	61200[2]	0.19[2]
	82800[48]	0.35[48]
Silver palladium alloy	80000[10]	0.33[10]
	95000[50]	0.35[50]
Lower modulus titanium	41360[22]	0.35[22]

The content of Table 4 is about material properties of composite resins which are supposed to tie between tissues and the materials used in implantology.

For analysis of the effect of the mastication act on mandibula or maxilla, soft, and hard foods have been modeled in a few studies.[25] Soft and hard food can be

Table 3. Elastic modulus and Poisson's ratio of auxiliary materials.

Material	Elastic modulus-E (MPa)	Poisson's ratio (ν)
Glass fiber	40000[26]	0.26[26]
Gutta percha	0.14[26]	0.32[35]
	0.69[35,61]	0.40[7]
	70[7]	0.45[26,61]
Carbon fiber	21000[26]	0.31[26]
Panavia cement	18600[26]	0.28[26]
RMGI adhesive	7600[15]	0.[15]
Plastic	2200[15]	0.30[15]
Zinc phosphate cement	17000[53]	0.35[53]
Elastic ligature	100[15]	0.30[15]

Table 4. Elastic modulus and Poisson's ratio of composite resins.

Material	Elastic modulus-E (MPa)	Poisson's ratio (ν)
RelyX Unicem 2 Automix Self Adhesive Cement Resin, 3M, EPSE, USA	15,8[67]	0.24[67]
True Vitality, Den-Mat Corp, Santa Maria, California	5.4[66]	0.24[66]
Herculite XRV, Kerr Corp, Orange, California	9.5[66]	0.24[66]
Charisma, Heraeus Kulzer, Hanau, Germany	14.1[66]	0.24[66]
Z100, 3M ESPE, St. Paul, Minnesota	21[66]	0.24[66]
Hybrid composite resin	22[12]	0.27[12]

modeled as an elastic boundary condition described by elastic modulus and Poisson ratios as $E = 2\,\mathrm{GPa}$ and $\nu = 0.4$, and $E = 10\,\mathrm{GPa}$ and $\nu = 0.28$, respectively.[25]

2.2.2. *Anisotropic, orthotropic, transversely isotropic, and nonlinear materials*

Due to the nonhomogeneous, anisotropic, and composite nature of bone, the mechanical responses obtained from compression and tension tests are different, namely the strain and stress results under the same loading are different with regards to the loading direction.[34] For instance, it was found by means of invasive techniques that the human mandible showed the stiffest characterization along the longitudinal direction.[77] The mechanical properties of bones obtained by noninvasive techniques (e.g., vibration response) can also be combined with the CT or MRI data and thereby more realistic models can be constructed.[78]

To acquire reliable numerical results from FEA, material properties of the corresponding biological and nonbiological structures should be modeled realistically. For example, O'Mahony *et al.* (2001) comparatively evaluated the mandible with completely isotropic and transversely isotropic models. They found that the transversely isotropic model provided higher stresses (up to 20%) than the isotropic

Table 5. Elastic modulus and Poisson's ratio of materials.

Material	E_x (GPa)	E_y (GPa)	E_z (GPa)	ν_{xy}	ν_{yz}	ν_{xz}	G_{xy} (GPa)	G_{yz} (GPa)	G_{xz} (GPa)
Enamel[12,71]	73.72	63.27	63.27	0.23	0.45	0.23	20.89	24.07	20.89
Dentine[12,71]	17.07	5.61	5.61	0.30	0.33	0.30	1.7	6	1.7
Cortical[20,35]	12.70	17.90	22.80	0.18	0.28	0.31	5	7.4	5.5
Cortical bone[2]	11.5	11.5	17	0.51	0.31	0.31	3.6	3.3	—
Cortical bone[53]	17.9	12.5	26.6	0.28	0.18	0.31	7.1	4.5	5.3
Cancellous bone[20,35]	0.21	1.148	1.148	0.055	0.322	0.055	0.068	0.434	0.068
Cancellous bone[53]	1.148	0.21	1.148	0.055	0.055	0.322	0.068	0.068	0.434
Fiber (in composite resin)[67]	37	9.5	9.5	0.34	0.27	0.34	3.544	1.456	3.544

x, y, and z represents the global coordinate system. ν is Poisson's ratio, E and G are elastic and shear modulus, respectively.

model at the crestal level.[44] They described the cortical bone as $E_x = E_y = 12.6$ GPa and $E_z = 19.4$ GPa, $G_{xy} = 4.85$ GPa and $G_{yz} = G_{xz} = 5.7$ GPa, $\nu_{xy} = \nu_{yx} = 0.3$, $\nu_{yz} = \nu_{xz} = 0.253$, $\nu_{\mathrm{zy}} = \nu_{zx} = 0.39$, and cancellous bone as $E_x = E_z = 1.148$ GPa and $E_y = 0.21$ GPa, $G_{xy} = G_{yz} = 0.068$ GPa and $G_{xz} = 0.434$ GPa, $\nu_{xy} = \nu_{zy} = 0.055$, $\nu_{yx} = \nu_{yz} = 0.01$ and $\nu_{xz} = \nu_{zx} = 0.322$, where the sub-indices x, y, and z denote global coordinate axes, ν is Poisson's ratio, E and G are elastic and shear modulus, respectively.[44,79]

In a different study, mechanical properties of various regions of condylar cartilage described as $E = 2.34$ MPa for anterior, $\nu = 0.46$; $E = 1.48$ MPa, $\nu = 0.39$ for central; $E = 1.51$ MPa, $\nu = 0.41$ for posterior; $E = 1.11$ MPa, $\nu = 0.38$ medial; and $E = 0.95$ MPa, $\nu = 0.31$ for lateral parts.[73]

Dentin and enamel also show anisotropic properties and regional stiffness alterations.[78] The data regarding the material properties of enamel and dentin could be gathered by microindentation techniques.[71]

In the case of the anisotropic modeling of biological tissues, various FEA studies in literature were briefly summarized in Table 5.

Discs act like a shock absorber. Structure of the disc can be characterized as a hyperelastic or viscoelastic model having a time-dependent response. The hyperelastic material is strengthened by fibers. The whole mechanical behavior of discs including fibers can be determined by experimental studies. However, since its isotropic character and the structural properties are not homogeneously distributed over the material, mechanical properties cannot be measured accurately in all points of a disc.[80] In general, hyperelastic models are described with the strain energy potential of the materials.[45,81] Then, stresses are obtained by deriving this potential energy relative to strain.

In the study of Koolstra and Van Eijden (2006), cartilaginous articular discs were described by Mooney Rivlin energy constants as $c_1 = 9.0 \times 10^5$, $c_2 = 9 \times 10^2$.[31,82] The same description was made for temporal and condylar cartilages as $c_1 = 4.5 \times 10^5$, $c_2 = 4.5 \times 10^2$.[31,82] Joisson *et al.* (2011) proposed in their study that the relationship between the Green–Lagrange deformation ε_{eq} and the Piola–Kirchhoff stress σ in a polynomial equation of the third order that can reproduce the hyperelastic behavior of the disc with Eq. (5). The values of the coefficients c_3, c_2, c_1, and c_0 are 104, 104, 0, and 1, respectively.[43]

$$\sigma = c_3\varepsilon_{eq}^3 + c_2\varepsilon_{eq}^2 + c_3\varepsilon_{eq} + c_0. \tag{5}$$

For the fluid part, the properties of water are described with density as $0.997\,\mathrm{g/cm^3}$ and viscosity as 8.899×10^{-4} kg/ms.[25] The fluid is accepted as laminar liquid (Reynolds number < 2100).[14]

2.2.3. *Mass density and yield strength value*

If the static case is taken into account in FEA, definitions of elastic modulus and Poisson's ratio would be enough to run such an analysis. On the other hand, if the dynamic case and/or vibration effect are investigated, mass density (ρ) of materials is necessary to be defined as seen in Eq. (6), where V is the volume of the model, a is the acceleration,t is the time, c is the damping factor, v is the velocity, k is the stiffness factor, x is the displacement, and F is the force. The mass density values used in the existing FEA studies in literature are given in Table 6.

$$\rho Va(t) + cv(t) + kx(t) = F(t). \tag{6}$$

Determination of the yield strength is a crucial step in the computational biomechanics, since beyond this strength value material would deform plastically. Once the yield point is exceeded, deformation would be nonreversible and permanent. Cansiz *et al.* (2015) assigned the values of yield strengths as 130 MPa for cortical bone, 462 MPa for titanium, and 72 MPa for resorbable materials.[47] Yield strengths of titanium, cobalt chromium, and feldspathic porcelain were determined

Table 6. Mass densities of various materials.

Materials	Mass density $\rho\left(\frac{g}{cm^3}\right)$
Enamel	2.9[25]
Periodontal ligament	1[25]
Cortical bone	1.3,[25] 1.5,[56] 2.4[30]
Cancellous bone	1.3,[25] 1.1[30]
Soft food	1[25]
Hard food	1[25]
Dentine	2.2,[67] 4[25]
Ni–Cr	8[67]
Composite resin	2.075[67]
Fiber	2.48[67]
Titanium	4.43,[67] 4.5[30]
Stainless steel	8[67]
Zinc phosphate	2.19[67]
Resin	2.19[67]

as 800, 720, and 500 MPa, respectively, in the study of Kayabasi *et al.* (2006).[2] According to Bayraktar *et al.* (2004), the yield strength of trabecular tissue is 135 MPa in compression, on the other hand, yield strengths of trabecular and cortical layers in tension are 85 and 108 MPa, respectively.[21,63]

As for the resorbable materials, Nieminen *et al.* (2008) reported the absorption time of a specific resorbable material, i.e., L-lactide, D-lactide, and trimethylene carbonate. This material can preserve its mass for 26 weeks within the human body. By the week 104, 63 to 80% of the mass will be absorbed. As for the resorbable screws, shear strength can be maintained for 12 to 16 weeks and this duration would be enough for the healing period of osteotomy.[83]

2.3. *Contacts*

Biological tissues have different types and numbers of connections to each other at the cellular level. Types of these connections mostly depend on the histological origin of the tissues within the mechanical and biological rules. Even though it is possible to identify a cellular-based link between anatomic structures, it is not possible to fully reflect the reality in FEA. On the other hand, generating a detailed and realistic model can cause errors occurring during the analysis phase.

The most frequently preferred type of contact in FEA is perfect bonded (fixed, rigid, preventing sliding, and separation at the interface) because of reducing computational cost and failure occurrence of the analysis process. The prefect bonded contact implies no slip interface between the components and strain values of adjacent surfaces would be very close to each other.[11,19,22,54,62,67,82] In order to acquire reliable numerical results, connections between anatomic structures and artificial materials should be defined in a more realistic fashion than a perfect bonded. Anatomic structures can make strict adaptations around artificial materials, but they cannot adhere these materials at cellular level even in osseointegrated materials.

In reality, most of the short- and long-term implant failures, such as decreased bone level and loss of osseointegration occur at the osseointegrated interface. In order to represent the nonosseointegration, bone, and implant interfaces should consist of the frictional surfaces. Smooth and porous or excessively rough surfaces should be described with lower and higher coefficients of friction, respectively.[84] Stress and also relative motion between surfaces are influenced by the value of the coefficient of friction. The surface to surface contact allowing small sliding without friction was constructed in some studies.[6] The friction coefficients between biological tissue and inorganic material, or between inorganic materials that were described in FEA studies in the literature are given in Table 7.

Geng *et al.* (2001) reported that the maximum cortical osseointegration (%100) over mandible is seen at anterior mandible.[34] The rate of osseointegration reduces toward the posterior mandible and the minimum value is observed at the posterior maxilla as less than 25%. The osseointegration depends on the stresses developed during healing, bone quality, and roughness on the surface of implant. In most of

Table 7. Friction coefficient used in FEA studies in literature.

Materials	Coefficient of friction
Titanium/cortical bone (implant/bone)	0.4[13,29]
	0.65[20]
Titanium/trabecular bone (implant/bone)	0.77[20]
	0.8[13]
	1[29]
Titanium/bone and callus	0.3[2,9,17,30,53,62]
Titanium/initial blood and granulation tissue	0.05[62,85]
Titanium/titanium (abutment/implant)	0.1[6]
	0.16[53]
	0.3[13,20,79]
	0.5[54]
Zirconia/ceramic (crown/veneer)	0.3[11]
Sandblasting titanium/bone	0.68[29,49]
Byplasma-spraying/porous beading techniques titanium/cortical	1[29]
Teeth/mandible	0.2[86]
Gold/titanium	0.2[53]
Zinc phosphate cement/titanium	0.2[53]
Zinc phosphate cement/gold	0.2[53]

the studies, in order to reduce the run-time of the analysis, %100 contact, namely perfect contact, between implant and bone is assumed.[34]

2.4. *Loading and boundary conditions*

Loading and boundary conditions, material and morphological properties, and porosity can affect the osseointegration. Especially loading and boundary conditions have a particular importance due to Wolff's law. According to this law, the bone subjected to loads will adapt to the new condition in such a way that the bone would improve its strength by remodeling itself to show resistance that loading. And if the loading on a particular bone decreases, bone atrophy occurs.[87]

2.4.1. *Loading conditions*

There are mainly three functional movements occurring during mastication process. These three movements i.e., biting, chewing, and clenching are not separate actions and occur jointly during mastication process. Temporomandibular joint (TMJ) is the most important anatomic structure that is responsible for this movement and it consists of complex substructures. These substructures are muscles (masseter, temporal, medial, and lateral pterygoid muscles), ligaments (temporomandibular, sphenomandibular, stylomandibular, and capsular ligaments) and articular disc that participate in the chewing action with other musculoskeletal anatomic structures of maxillofacial region. Although, TMJ and its main components dominate the chewing action, it is a complex movement involving all the musculoskeletal elements of the maxillofacial region which makes the 3D modeling of this structure complicated.

Table 8. Static and dynamic unidirectional loads.

Area of load application	Load and direction
On metal support of prosthesis[11]	1 N vertical
Top surface of the abutment[13]	190 N at 30° relative to the long axis of the implant
Maxilla, occlusal force, on the top of the implant abutment[62]	150 N vertical load
On the posterior teeth to simulate maximum occlusal force[8]	300 N vertical load
The central pit[28]	300 N vertical load
Clenching force, molar region at the same site of fracture line[46]	62.8 N vertical to the occlusal plane
Masseter muscles area[86]	300 N in direction of masseter muscles
Incisor bite force[51,89]	62.8 N vertical load
On teeth[30]	100 N Axial, 1 Hz
On the lower first molar[21]	132 N occlusal load
Palatal side of the abutment[64]	178 N perpendicular to the palatal surface

On the other hand, TMJ has multi-directional action. Bilateral translation and rotation movements of the TMJ move the mandible on the sagittal and axial planes. In addition, unilateral movements of TMJ can move the mandible on the coronal plane. This ability of TMJ enables mandible to move in three different planes in space delimited by the ligaments spanning the joint.

An adult has the capability of applying a bite force between 244 and 2143 N.[8] The highest occlusal biting force is between 297 and 669 N and between 42 and 412 N for natural teeth and implant, respectively.[9] The maximum loading is between 244 and 1243 N for full dentition while clenching.[88] Gross *et al.* reported from literature survey that bite force ranged between 383 and 880 N and 176 and 229 N for molars and incisors, respectively.[56] Mean swallowing force of 665 N and chewing of 267 N have also been reported.[56] Hsu *et al.* stated that the average bite forces in the molar region were measured more than 600 N for female young adults and 800 N for male young adults.[57] The used unidirectional load value, direction and area of load application in FEA studies are listed in Table 8.

Generally, the static loads taken into account in FEA are the masticatory forces. In the studies including static loadings, oblique occlusal forces are required to be considered to obtain a realistic model (Table 9). Since the fatigue failures occur due the repetitive dynamic loads, the effects of dynamic forces deserve to be carefully investigated.[34] If the static loading condition is replaced with the same magnitude of dynamic loading, the maximum stress values observed over the material can increase up to 20%.[34,90]

In literature, a few studies used the muscle forces that were obtained from either experimental study or computational process.[6,17] The muscle forces applied at insertions on the bone as a vector quantity are summarized in Table 10 with detail. The table contains the information of amplitude and direction of muscle forces. The muscle loads were applied on pre-determined insertion points on the mandible.

Table 9. Static and dynamic multidirectional loads.

Area of load application	Amplitude and direction of the load
On both cusps[12]	100 N palatal load in the palatal cusp
	100 N buccal load buccal cusp at 45° to the long axis
On the insertion area of the masseter muscle on the zygomatic arch[5]	300 N distributed force, direction of muscle
	150 N vertical load
On the implant[9]	300 N vertical load to long axis of implant
	150 N top of the abutment at an inclination of 60° buccally from the vertical
To the occlusal plane of implant[42]	114.6 N axial
	17.1 N lingual
	23.4 N disto-mesial
Occlusal surface of the crown[29]	200 N vertical
	20 N lateral
On the crown[35]	200 N vertical intrusion along the axis of implant
	100 N shear force opposed to the buccal–lingual axis
On the implants[2]	118.2 N 75° to the occlusal plane
	Dynamic; 75° to the occlusal plane 118.2 N at 0.08 Hz during 5 s
	Dynamic; simulated the draught of a hot (60°C) and a cold (15°C) liquid for 1 s.
Most coronal part of the crown[44]	50 N vertical through the long axis of the restoration and implant
	10 N horizontal from buccal to lingual
On both cusp[10]	50 N vertical on buccal cusp
	50 N vertical on lingual cusp
On the top surface of the abutment[20]	150 N along the implant long axis
	150 N abutment long axis
On the abutment[54]	100 N vertical
	100 N 15°, 30°, and 60° to the vertical axis
Buccal cusp tip[28]	50 N at 45° tooth axis
On the implant[22]	0–150 N axial (dynamic, 5 s, 0.08 Hz)
	0–25 N 45° to the occlusal plane (5 s, 0.08 Hz)
On the implant[88]	100 N vertical
	30 N 45° to the occlusal surface
Top surface of abutment[79]	170 N vertical
	170 N 45° to the long axis of implant
On the crown[48]	150 N vertical to the tip of buccal cusp
	150 N vertical to the distal fossa
On the crown[48]	100 N vertical to the tip of buccal cusp
	100 N vertical to the distal fossa
	100 N vertical to the mesial fossa
The incisal edge of the pontic[71]	154 N at a 45° angle
On the abutment[53]	35 N cm torque to screw head
	100 N vertical to occlusal area of teeth
	100 N 45° in the buccolingual direction to occlusal area of teeth
On palatal surface of the crown[91]	25.5 N Horizontal 90° to long axis of implant
	178 N Oblique 30° to long axis of implant
To the centric stop of the occlusal surface In centric occlusion[26]	400 N at a 45° inclination in a linguolabial direction to the long axis of the tooth

Table 9. (*Continued*)

Area of load application	Amplitude and direction of the load
At the buccal cusp tip[28]	100 N parallel or 45° to the long axis of the tooth
At the pit in the central buccolingual section of the model[28]	135 N parallel or 45° to the long axis of the tooth
On the buccal cusp tip[28]	250 N at 40° oblique angle
On the middle of the top surface of the angled abutment of anterior teeth[49]	178 N 120° to the abutment's long axis
The load applied near the cingulum area of the prosthesis[52]	178 N 130° to the long axis of the implant
3 mm below the incisal edge on the palatal surface[50]	146 N 135° to the long axis of prosthetic-driven model
The center of the palatoincisal line angle of abutment[19]	176 N 120° to the implant long axis

Table 10. Components of maximum forces of some masticatory muscles along mediolateral, apicocoronal, and anteroposterior axes that activate jaw for closing movement.[6,17]

Muscles	Force (N) (mediolateral)*	Force (N) (apicocoronal)*	Force (N) (anteroposterior)*
Superficial masseter	7.78	127.23	22.68
Deep masseter	12.87	183.50	12.11
Medial pterygoid	140.38	237.80	−77.30
Anterior temporalis	0.06	0.37	−0.13
Medial temporalis	0.97	5.68	−7.44

Note: *Axis is positioned between condyles.

Table 11. Masticatory muscle forces, directions and lengths.[36,75]

Muscles	Direction cosines of resultant muscular forces			Muscle length (mm)	Magnitude of forces developed by muscles (N)
	1	2	3		
Right side					
Posterior temporalis	−0.1	0.76	0.64	62.9	20.2
Anterior temporalis	−0.07	−0.34	0.94	57.4	21.9
Internal pterygoid	0.32	−0.03	0.94	43.3	16.1
Superficial masseter	0.27	−0.15	0.95	48	9.8
Internal masseter	0.27	0.18	0.94	29.5	16.1
Left side					
Posterior temporalis	0.1	0.76	0.64	62.9	11
Anterior temporalis	0.07	−0.34	0.94	57.4	27.9
Internal pterygoid	−0.32	−0.03	0.94	43.3	17.1
Superficial masseter	−0.27	−0.15	0.95	48	20.2
Internal masseter	−0.27	0.18	0.94	29.5	27.3

Table 12. Maximum forces of some mandibular muscles.[93]

Muscles	Maximum force (N)
Superficial masseter	190.4
Deep masseter	81.6
Medial pterygoid	174.8
Anterior temporalis	158
Middle temporalis	95.6
Posterior temporalis	75.6
Inferior lateral pterygoid	66.9
Superior lateral pterygoid	28.7
Anterior digastric	40.0
Anterior mylohyoid	20.0
Posterior mylohyoid	20.0

Table 11 includes muscle lengths, direction cosines of muscle fibers and magnitude of muscle forces included in FEA studies. Origin and insertion points of muscles were defined and both of them were coupled.

For detailed muscle information, readers are referred to Refs. 37, 80, and 92. Van Eijden *et al.* (1997) measured physiological parameters of some of the mandibular muscles such as pennation angle, fiber length, and cross-sectional area from cadavers, which affect the muscle force production capability.[80] Posterior and anterior temporalis, internal pterygoid, superficial, and internal masseter muscles are especially active during the mastication process. Digastric, geniohyoid, and mylohyoid are the other mandibular muscles and they are particularly active during jaw opening. Maximum forces of some mandibular muscles are given in Table 12, which are drawn from Ref. 93.

In the distinctive study of Gross *et al.* (2001), rod elements were preferred to model active muscles during the mastication process by connection to the skull.[56] The distances between the origins and insertions of the muscles were equal to the length of the rods. Rod thicknesses were proportionally determined according to the cross-sections of the muscles. Data on muscles' cross section areas were derived from the literature. The Young's modulus was chosen to be 0.8 MPa. The number of rods were distributed muscles by muscles for example, 15 rods, determined each one was $0.35\,\mathrm{cm}^2$ for temporalis muscles; four rods, determined each one was $1.33\,\mathrm{cm}^2$ for masseter muscle; four rods, determined each one was $0.94\,\mathrm{cm}^2$ for medial pterygoid muscles; and four rods, determined each one was $0.75\,\mathrm{cm}^2$ for lateral pterygoid muscles.[56] Although the determined muscular forces by a linear relation between the force and cross-sectional area of muscle is not an accurate method, it is still the most convenient approach for muscle modeling in FEA.

2.4.2. *Boundary conditions*

Maxilla consists of two different fused parts and it articulates with nine different bones. These bones are nasal, zygomatic, palatine, lacrimal, vomer, inferior nasal concha, frontal, and ethmoid bones and they constitute skeletal bony structure of the face and anterior head. In addition, maxilla may articulate with orbital floor and lateral perpendicular plate of sphenoid bone. These articulations are provided with suture called joint surfaces and these surfaces are very active during adolescence. After this phase, they are completely ossified and became mechanically durable.

During definition of the boundary conditions in the finite element method, all of the sutures may be separately defined to isolate maxilla or the whole cranium or to model as a unique solid part. Although modeling the cranium as a solid one-piece structure is easy to design, the presence of the whole cranium, instead of the isolated maxilla model, may cause analysis failures. On the other hand, using isolated maxilla models may result in a more realistic way for the evaluation of maxillary biomechanics behaviors. The temporal surface and incisor point can be fixed.[6]

For the mandible, the boundary conditions are described easier than maxilla. The movements of the mandible and condyles during biting can be assumed to be restricted in translational movement and only rotational movements are allowed. Therefore in most studies, tops of two condyles are fixed in space by selecting a condyle head area or defining reference points coupled to condyle for acting same displacements.[32,36,38,47,89] In reality, translational motions of the condylar heads along the mediolateral and anterior–posterior axes, as well as the rotational motion, should be taken into account.[94]

In Ref. 94, load and boundary conditions were described in two steps. First, mandibles were positioned such that to be close in occlusal plane of molar region without force. Then, loads were created by dominant muscles when the mastication on mandible was performed. The interval between maxilla and mandibula measured 5 mm from CT of ten volunteers on average was described by a boundary condition, which was applied on the molar region at the same site of the fracture line. The condition, arising from position of maxilla, was simulated close position of mandible. In addition, the rotational movements of condyles were allowed, the vertical translations of condyles were restricted 5 mm on average, the anterior–posterior direction of condyles was restricted 4 mm in average, and the lateral translate was kept fixed. Dimensions of boundary conditions were acquired from CT images. The boundary conditions were the same in each analysis step.[94]

Generally in FE studies, bone block or only PDL and teeth are modeled to reduce the computation duration. If the model has got teeth, PDL and prosthesis, external surfaces of PDL should be restricted in all translational movement.[7,11] If the model has been comprised of bone block, the bone mesial and distal surfaces can be constrained in all direction.[9,13,19,20,25,30,79] Some studies preferred to restrict bone block from lateral surfaces of base of cortical and trabecular bone,[12,64] entire outer bone[26,54] and buccal and lingual fraction of cortical bone.[29,35]

2.5. *Elements*

Element type should be chosen considering the morphology and, stress and strain degree. Element properties can be chosen according to the following criteria. Beam elements are used, if one direction of model is longer and the content of the study is focusing on the bending moment: Plane stress elements are used, if the stress vector is zero at particular surface (e.g., soft tissue). Plane strain elements are used, if the strain tensor is zero at particular direction (e.g., soft tissue). Shell elements are used, if the model is thin but has 3D geometry such as mandible and cement. All 3D elements i.e., solid elements are used for calculating the stresses along all direction, but much computational time is required. In general, 2D plane stress or plane strain, shell, rod, truss, 3D shell or solid element, and beam elements family can be defined for tissues and artificial models.[78]

Elements can be categorized by considering the movement ability of the construction i.e., degrees of freedom (DoF). Linear (first order), parabolic (second order), and quadratic (third order) elements with two, three, and four nodes along each edge are used, respectively. By increasing the level of order, the element becomes much more flexible because of increasing the DoF of element, so the model is obtained with more accurately. Element topology depends on shapes of the element such as triangle and quad in addition to dimension of the element such as 1D, 2D, and 3D. Although the quad element obtains the result more accurately than triangular one due to higher DoF number, triangular element is preferred due to easier fit on rough and complex surface of bone.[78]

In most of the studies, meshes comprise 10 node 3D tetrahedral quadratic elements.[11–13,16,22,35,79] To make a decision about the distance between two nodes on the same element is important for getting accurate results. In literature, distances between two nodes of a tetrahedral element are 0.2,[11] 0.25,[82] 0.35,[35] 0.4,[13] 0.5,[20,79] and 0.75 mm.[46] Eight node 3D hexahedral linear element is used without the information about distance.[9,49,50,67] For structures in large strain, the 3D shell element was suggested by Sussman and Bathe (2012).[95] The cement was modeled by 3D shell element in certain studies.[46,62,67] Four nodes with three DoF per node, tetrahedral linear elements were used and 0.2 mm distance between two nodes was described in some studies[30,36,64] and the distance of node for critical area was meshed as 0.02 mm.[42]

Even though most of the studies stated that the accurate result can be calculated with the highest number of elements, the optimum length is a better choice for accuracy and reducing time consumption of analysis processes. H analysis provides learning optimum element size and type. Stress data is compared on same point of bone or implant for each model which has got different size of element. The graph should be lined with two axes as different element size and maximum stress on the same point. At the end, the optimum number of element can be decided when the maximum stress does not change size by size. If the analysis has got too small element size, time consumption increases and the result would not be accurate. The effect of Saint Venant causes the virtual stress concentration on the loading point for some area regarding loading and boundary conditions.

The cable element for muscles and ligaments and the compressive gap element for discs can also be used. Furthermore, models with nonlinear spring elements are also a widely preferred method for ligaments. As it can be deduced from the study of Chen and Xu (1994), each ligament has proportional numbers of spring depending on their cross sectional area obtained from four human TMJs. One of the springs has a cross-sectional area of $0.25\,\mathrm{mm}^2$ and a stiffness of $10.9\,\mathrm{N/mm/mm}^2$. The upper bands of the posterior capsular ligament has six springs, the lower bands of the posterior capsular ligament has five, and the anterior capsular ligament has four, proportionally.[74] In another study, the truss element and membrane element were used for four elevator muscular groups as masseters and medial pterygoids and temporalis and lateral pterygoids, respectively.[76]

3. Future Research Directions

FEA consists of different steps such as (i) 3D data collection, (ii) 3D model creation, (iii) mesh processing, (iv) determination of boundary conditions, (v) determination of the type of the analysis, and (vi) the application of the forces. Some of these processes are performed via different specific software and 3D modeling requires a kind of data producing system, such as CT or MRI. In the future, a compact system, which has the ability to make all these processes in the same unit, may be designed and hence current operating systems would be improved. When specific FE models of subjects from CT images can be analyzed in a single system without consuming much time, an optimum customized implant can be practically designed and produced for each patient. Modeling and material of the implant or screw may be determined for each bone and each fracturing type, individually.

Soft tissues such as muscles, ligaments and disc can be modeled from MRI data due to their better resolution than CT. Thus, volumetric model of soft tissue can be obtained in 3D space. The 3D soft tissue model prevents the virtual stress concentration on the attached point of the bone. Nevertheless, the model of healthy disc may be suggested as fluid filled cavity.

In order to obtain reliable FEA results, material properties of bone, muscle, ligament and inorganic structure should be determined accurately i.e., modeling with nonhomogenous, anisotropic, and nonlinear material properties.[96,97] Nonhomogeneous structure can be obtained from CT data and a model can be produced by software such as Mimics or Bonemat.[39,41,96] Any technique for determining anisotropic structure has not been searched out yet. A noninvasive method that is able to acquire stress values in all directions in vivo and in real time would be very efficient in the future. But today, for describing anisotropic properties, data should be gathered from invasive experimental study and these data cannot be subject specific.

The FE model should also consist of the bone remodeling process with regards to stress distribution due to Wolff's law. When applied load on the specific parts of a bone increases, value of the material properties also increases or porosity of bone decreases. If the load application decreases, the bone becomes less dense and value of material properties decreases or porosity of bone increases like osteoporosis problem.

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