Biomechanical Analysis of Stress and Stiffness of New Load-Bearing Implants

JINXING HUO
Abstract

Medical implants are essential products for saving lives and improving life quality. Nowadays, the demand for implants, especially biocompatible and personalized ones, is increasing rapidly to deal with factors like congenital malformations, aging, and increasing prevalence of cancer. To facilitate their clinical applications, better understanding of their biomechanical properties is important. This thesis focuses on tubular and mandibular implants, and aims at studying stiffness properties and assessing stress distributions.

Tubular implants with coupled helical-coil structure, which can be potentially used as tubular organ constructs, were manufactured by winding polycaprolactone filaments. Tensile and bending stiffnesses were evaluated through mechanical testing and finite element simulations. By increasing the number of helical coils, we could realize a new type of tubular implants which could be used in applications like trachea and urethra stents. Stiffness properties of such implants were investigated analytically, due to the geometrical periodicity. Through computational homogenization, the discrete mesh structures were converted to equivalent continua, whose structural properties were studied using composite beam theories. The numerical and analytical models developed can serve as tools for the mechanical design of implants.

A patient-specific mandibular implant, additively manufactured of titanium alloys, failed shortly after surgery. The failure was studied using a numerical approach. Finite element models were generated from the 3D bone reconstructed from computed tomography data and implants processed by computational homogenization. The failure location and that of the numerically predicted largest von Mises stress agree well, which confirms the feasibility of using finite element simulations to quantitatively analyze implant failures and assist in implants design.

For implant failures caused by local bone loss, analytical studies were also carried out to assess the stress distribution around screw-loaded holes in bones. The mandibular bone was treated as a laminate of which elastic properties were obtained by classical laminate theory. The stress profiles were predicted using a complex stress function method. The loading direction was found to have a minor influence on the stress distributions, while the friction coefficient has non-negligible influence. The stress state can serve as starting point to predict bone remodeling and be compared with criteria for bone strength.


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ISSN 1651-6214
ISBN 978-91-554-9342-4
urn:nbn:se:uu:diva-262688 (http://urn.kb.se/resolve?urn=urn:nbn:se:uu:diva-262688)
List of Papers

This thesis is based on the following papers, which are referred to in the text by their Roman numerals.


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The following contributions have been made during the PhD period but are not included in the thesis.

**Journal articles**


**Conference contributions**


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# Abbreviations

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<tr>
<td>AM</td>
<td>Additive manufacturing</td>
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<tr>
<td>CAD</td>
<td>Computer-aided design</td>
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<td>CNC</td>
<td>Computer numerical control</td>
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<td>CT</td>
<td>Computed tomography</td>
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<td>EBM</td>
<td>Electron beam melting</td>
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<td>FE</td>
<td>Finite element</td>
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<td>RVE</td>
<td>Representative volume element</td>
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<td>PCL</td>
<td>Polycaprolactone</td>
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1. Introduction

1.1 Medical implants

Medical implants are defined, in a medical context, as devices or tissues that are placed inside or on the surface of the body to replace missing biological structures, support damaged structures or enhance existing structures. The use of medical implants dates back to the time of Egyptians, with the oldest known prosthesis being two toes belonging to Egyptian mummies [1]. Major implants came into use during the 16th century due to the increased desire by doctors to treat amputated sailors having sustained injuries at sea or soldiers wounded on the battlefields. Such surgical treatments at that time had low degree of success partly due to the lack of precision in the manufacturing of medical implants. In the 20th century, a profound development in the field of medical implants took place with many breakthroughs in both procedures and materials used in prosthetics, such as rapid prototyping, computer-aided design, resorbable plastics and titanium alloys. Recently, additive manufacturing (AM) technologies have opened a new window for making anatomically shaped implants [2–4]. Customized net-shape implants can be produced layer by layer to allow well-controlled and interconnected porosity which, combined with solid material, provides better bone ingrowth into implants [5].

Nowadays, medical implants are being used in many parts of human bodies for various applications like orthopaedics, pacemakers, cardiovascular stents, defibrillators, neural prosthetics and drug delivery systems [6]. There is no doubt that implants have improved the medical outcomes for millions of patients and will benefit even more in the future [7]. The development of medical implants is a complex, multi-stage design and manufacturing process. Generally, the development process can be summarized as the following six steps [8]:

1. Definition of the problem based upon needs and objectives of the working environments of the implants;
2. Preliminary ideas for implants are formulated and preliminary design is created on the basis of radiographic data;
3. Prototype production and improvement based on theoretical analyses;
4. Verify the performance and functionality of the prototype with different mechanical, chemical, histological and cadaver tests;
5. Further in-vivo tests of the prototype on patients;
6. Clinical use of the developed implant.

1.2 Motivation and aim of the study

Bone fractures, resulted mainly from impact of large loads or cyclic activity of small loads, are observed fairly frequently nowadays [9]. As reported, there were a total of about 9 million fractures occurring in the year 2000, including 1.4 million hip fractures, 1.7 million forearm fractures and 1.4 million vertebral fractures [10]. The need of medical implants is skyrocketing due to medical treatments for the aforementioned bone fractures and some other factors, such as congenital malformations, aging, and increased cases of cancers. For instance, with the life expectancy rising worldwide, the number of older people who need replacement of failed tissue, e.g. hip, with implants made of biomaterials is increasing.

The huge demand for implants leads to significantly increased cases of surgeries performed each year for transplantation and also drives advancement in the fields of biomaterials, manufacturing methods and computer technologies. Thus, more tailored implants with better biomechanical properties, such as biodegradable and patient-specific implants are being fabricated and demonstrate high potential for clinical use. However, the understanding of the structural and mechanical properties of new implants is still insufficient. This slows down the process of getting new implants into practical use. Moreover, complex-shape implants made by additive manufacturing, with design much dependent on the individual experience of surgeons, might lead to slow and expensive final results or even postsurgical implant failures.

For the purpose of facilitating the application of new implants and improving patients’ quality of life, this thesis is devoted to the study of new types of medical implants from a biomechanical perspective. Papers I-III deal with tubular implants, and Papers IV and V deal with mandible implants.

In Paper I, the aim was to manufacture a tubular implant with coupled helical structures and to develop a finite element (FE) model for coupled helical coils for study of the structural response under small-load conditions. In Paper II, the aim was to find an equivalent continuum description of a mesh structure with a rhombic representative volume element. In Paper III, the aim was to develop closed-form solutions for assessing the stiffness properties of mesh tubes. In Paper IV, the aim was to investigate the fracture of a mandibular implant from a clinical case in order to explore the capability of FE analysis. In Paper V, finally, the aim was to predict the stress distribution around a screw-loaded hole in a human mandible by utilizing an analytic model for a pin-loaded plate.
2. Implant biomechanics

The word "biomechanics" is coined from the ancient Greek words βίος ("life") and μηχανική ("mechanics") and it refers to the study of the movement of living things using the science of mechanics from the anatomical and functional aspects [9, 10] as indicated by Figure 1. In this thesis, the focus is on studying biomechanical problems concerning human beings. Biomechanics provides conceptual and mathematical tools that are essential for understanding how humans move and how kinesiology professionals might improve movement or make movement safer [13].

Figure 1. Biomechanics uses the principles of mechanics for solving problems related to the structure and function of living organisms [12].

Biomechanics can be subdivided into three distinct areas: sports biomechanics, occupational biomechanics and clinical biomechanics. Clinical biomechanics refers to development of treatment protocols and methods to assist in medical treatments increasing the quality of life of individuals suffering from a health conditions, disease or illness inhibiting their ability to move and function properly. Among the various applications of biomechanics, one oriented towards medical implants is in the focus of this thesis. Implant biomechanics is defined as the study of mechanical properties of implants or prostheses using the principles of engineering mechanics.
2.1 Tubular implants

Tubular organs of human beings refer to organs with a tubular structure, such as the gastrointestinal tract, urinary tract and vascular system. In the case of tumor or trauma in the sites of such organs, the corresponding surgical treatment generally necessitates tissue-engineered tubular implants. For example, respiratory system diseases, like tracheal damages due to cancer or congenital diseases, are one of the most common causes of death. Implants are necessary for curing such problems if the damaged area is beyond the limits of end-to-end anastomosis. One type of trachea implant, demonstrated in Figure 2(a), could be used to restore the structure and function of severely damaged airways. Moreover, for patients suffering from coronary heart disease caused by constricted arteries, a grid-like coronary artery stent, as shown in Figure 2(b), can be inserted through the constricted area to open the narrowed arteries and help reduce symptoms such as chest pain or to help treat a heart attack. Similar treatment also applies to patients with benign esophageal stricture where the esophagus is narrowed by scar tissue.

![Figure 2. Examples of tubular organ implants. (a) Trachea implant (Harvard Apparatus Regenerative Technology Inc.) [14] and (b) coronary artery stent (Centrele de Excelenta ARES) [15].](image)

2.2 Mandible implants

The human mandible, i.e. the lower jaw, is the strongest bone of the face. It is composed of a horizontal horseshoe-like body, of which the ends are connected to two perpendicular portions, the rami, at virtually right angles. The main function of the mandible is to support the lower part of the face and to keep the lower teeth in place.

Malfunction of the mandible due to, e.g. tumors, bone infections and the like, leads to severe deformation and difficulty in chewing. Therefore, medical measures should be taken to retain its functionality. Generally, such measures involve treatments to surgically resect a certain section of the mandible, e.g. around the tumor, and to reconstruct this section with implants, e.g. a patient-specific mandible, as shown in Figure 3(a). Another main cause leading to surgical treatment is mandible traumas, which may
result in one of the most common skeletal injuries in human, mandible fractures. The surgical treatment of severe mandible fractures is rigid fixation using different types of implants, like compression plates, reconstruction plates and mini plates, as shown in Figure 3(b).

Mandible implant failure almost always occurs at the plate bending area [16] and usually results in unsatisfactory function. Understanding the nature of the loading forces, the mechanism of transfer of such loads to the bones, and distributions in implants has been recommended as important factors to be considered in the development of new mandible implants and, achievement of implant longevity.

Figure 3. (a) Patient-specific mandible implant (3DSystems) [17] and (b) Thread-Lock TS SmartPlate (KLS Martin) [18].
3. Methods and materials

3.1 Finite element method

The era of the FE method began in the 1940s in the field of solid mechanics. Several literatures [19–21] studied the stresses in continuous solids by replacing a small portion of a continuum with a lattice of line. Later in 1956, Turner et al. proposed a more direct substitution of properties which was more effective by assuming that all elements behave in a simplified manner. In 1960, Clough first coined the term “finite element”. Up to the early 1960s, most FE studies were focused on small strains and small displacements, elastic material behavior and static loadings. Later on, the finite element method was expanded to take into account large deflection, thermal analysis, material nonlinearities, buckling problems and visco-elasticity problems by Turner et al., Gallagher et al., and Zienkiewicz et al., [22–24].

The FE method could be regarded as a discretization of continuous physical model into finite elements and governing laws are applied on individual element, as shown in Figure 4. Within a discrete element, defined by the mesh of the model, a field quantity, e.g. displacement, is interpolated from nodal values of this quantity. When all elements are connected, the field quantity becomes interpolated over the entire structure in a piecewise way by polynomial expressions of which the number is the same as that of elements. The solutions of the nodal values of the field quantity are determined by minimizing certain function such as the potential energy.

\[ \varepsilon_{ij} = \frac{1}{2} \left( \frac{\partial u_i}{\partial x_j} + \frac{\partial u_j}{\partial x_i} \right) \]

\[ \sigma_{ij} = D_{ijkl} \varepsilon_{kl} \]

\[ \frac{\partial \sigma_{ij}}{\partial x_i} + \frac{\partial \sigma_{ij}}{\partial x_j} + \frac{\partial \sigma_{ij}}{\partial x_3} + \bar{f}_i = 0 \]

Figure 4. General procedure of FE method.
Favoured by the evolution of computer technology, FE analysis has gained extensive attention from a relatively new application field – biomechanics [25–28]. In musculoskeletal research, e.g., the FE method is used to deal with the kinematics of the skeletal system and muscle activity [29–31]. Stress and strain analyses of bone, joints and load-bearing implants are also implemented through FE modeling [23, 24]. Moreover, the FE method can be used for implant development [34], helping to answer unresolved questions related to clinical complications. The FE method has been used in Papers I and III to model the structural responses of coupled helical coils, in Paper II to assist in computational homogenization of mesh structures, and in Paper IV to study the failure of a mandible implant.

3.2 Computational homogenization

Multi-scale methods, aimed at predicting collective multi-phase response of materials, can be classified in various ways. One category is homogenization methods based on integration over small scales. Among these methods, computational homogenization provides one of the most accurate techniques for upscaling the nonlinear behavior of a well-characterized microstructure [35]. The idea of computational homogenization was presented by Suquet [36] and then further improved by other scholars in the early 21th century [37–39].

In Paper III we used first-order computational homogenization which indicates that the response at a macroscopic material point is determined only by the first gradient of the displacement field [40]. This method is illustrated in Figure 5, and its main principle can be described by four steps as follows [36]:

- Creation of a microscopic representative volume element (RVE);
- Definition of the microscopic boundary conditions based on macroscopic inputs. In this thesis, the RVE deformation is chosen to be driven by the macroscopic deformation gradient;
- Obtaining the macroscopic stress tensor, by solving the boundary value problem on the RVE level;
- Determining the constitutive tangent form of the relation between macroscopic inputs and outputs.

The mathematics behind first-order computational homogenization is averaging theorems which couple the macroscopic and microscopic scales. These theorems establishes that the variation of potential energy at a macroscopic material point is equal to that in a RVE [41]. A commonly used theorem is the Hill-Mandel principle of macro-homogeneity [42]. The macro-micro coupling is implemented by imposing the deformation gradient in a
certain way. Three frequently used ways are prescribed displacement, prescribed traction and periodic boundary conditions [43].

3.3 Image analysis techniques

3.3.1 Metallic artifacts deletion

Since its introduction in the 1970s, computed tomography (CT) has become an important medical imaging tool for diagnostic use. Till now, metal artifacts, which cause shadows and streaks in the reconstructed images, have been a disturbing problem. Prior to analysis of CT images, contaminated slices need to be cleaned up by use of metal artifact reduction techniques. Among available methods [44–46], an iterative method called Metal Deletion Technique (MDT) was employed in Paper IV. This technique is based on the circumstance that projection data involving metal or near metal are less accurate [47]. The basic algorithm of MDT is depicted in Figure 6. Forward projection is performed in an iterative way to replace the detected measurements involving metal.

3.3.2 Segmentation

Due to the fast developments of computer technology, research combining imaging techniques like CT with the FE method is emerging in the field of biomechanics [48]. A key procedure for linking CT data and FE analysis is the reconstruction of tissue geometries from CT slices, which is frequently accomplished by the process of segmentation. Among the available methods, thresholding-based techniques have been widely used [49]. Image thresholding methods are normally categorized into two classes: global thresholding
and local thresholding. In the analysis of hard tissues like bone, global thresholding is often chosen since bone structures are of higher intensity levels than soft tissues like muscles [51]. It was used in Paper IV. The principle of global thresholding is expressed by the function [52]

\[ g(x, y) = \begin{cases} 1 & \text{if } I(x, y) > T \\ 0 & \text{if } I(x, y) < T \end{cases} \]  

in which \( g \) is the value of the pixel \((x, y)\) in the resulting image, \( I \) is the image intensity at the pixel, and \( T \) is the prescribed threshold value. The result is a binary image, in which pixels with intensity one represent the potential region and pixels with intensity zero represent the background.

### 3.4 Classical laminate theory

In Paper V, classical laminate theory has been used to calculate the material properties of the mandible bone. Analogous to Euler-Bernoulli beam theory and classical plate theory, classical laminate theory is valid only for thin laminates with small displacement in the out-of-plane direction. Except for the perfect-bonding assumption, the basic assumptions of this theory are the same as the Kirchhoff hypotheses in classical plate theory [53]:

- Straight lines normal to the mid-surface remain straight;
- Straight lines normal to the mid-surface remain unstretched;
- Straight lines normal to the mid-surface remain normal.
The laminate, which is assumed to be homogeneous and not necessarily isotropic, is subjected to both transverse and in-plane loadings. The applied load is determined by the relation

\[
\begin{bmatrix} \mathbf{N} \\ \mathbf{M} \end{bmatrix} = \begin{bmatrix} \mathbf{A} & \mathbf{B} \\ \mathbf{B} & \mathbf{D} \end{bmatrix} \begin{bmatrix} \mathbf{\epsilon}^0 \end{bmatrix},
\]

(2)

where \( \mathbf{N} \) is the force vector, \( \mathbf{M} \) is moment vector, \( \mathbf{A} \) is the extensional stiffness matrix, \( \mathbf{D} \) is the bending stiffness matrix, \( \mathbf{B} \) is the coupling stiffness matrix, \( \mathbf{\epsilon}^0 \) is the strain vector, and \( \mathbf{k} \) is the curvature vector. The components of these stiffness matrices are

\[
A_{ij} = \sum_{k=1}^{n} (C_{ij})_k (z_k - z_{k-1}),
\]

\[
B_{ij} = \frac{1}{2} \sum_{k=1}^{n} (C_{ij})_k (z_k^2 - z_{k-1}^2),
\]

\[
D_{ij} = \frac{1}{3} \sum_{k=1}^{n} (C_{ij})_k (z_k^3 - z_{k-1}^3),
\]

(3)

where \( z_{k-1} \) and \( z_k \) are the distances from the bottom and top surfaces of the \( k \)th layer to the mid-plane, respectively, \( C_{ij} \) are the components of the stiffness matrix of each layer, and \( n \) is the total number of layers.

When only in-plane loadings are applied to the laminate, it is often convenient to use what can be referred to as effective engineering properties. In the case of a symmetric laminate, the components of \( \mathbf{B} \) are identically zero. Thus, the coupled relations in Eq. (2) decouple into two independent equations. The one which represents the relation between external forces and laminate strains is given by

\[
\begin{bmatrix} N_x \\ N_y \\ N_{xy} \end{bmatrix} = \begin{bmatrix} A_{11} & A_{12} & A_{16} \\ A_{12} & A_{22} & A_{26} \\ A_{16} & A_{26} & A_{66} \end{bmatrix} \begin{bmatrix} \mathbf{\epsilon}^0_x \\ \mathbf{\epsilon}^0_y \\ \gamma_{xy} \end{bmatrix}.
\]

(4)

By introducing average quantities with respect to the thickness of the laminate, one can derive from Eq. (4) the effective elastic moduli, the shear modulus and the Poisson’s ratio

\[
\bar{E}_x = \frac{1}{a_{11}H}, \quad \bar{E}_y = \frac{1}{a_{22}H}, \quad \bar{G}_{xy} = \frac{1}{a_{66}H}, \quad \bar{v}_{xy} = \frac{a_{12}}{a_{11}},
\]

(5)
where $a_{ij}$ are the components of the inverse matrix $A^{-1}$ and $H$ is the thickness of the laminate.

### 3.5 Complex stress function method

Analytical solutions of the stress distribution in orthotropic homogeneous plates with circular openings filled by elastic material with different elastic properties have been derived by Lekhnitskii [54]. When the opening (far from the edge) is relatively small in comparison with the plate size, the plate can be simplified as infinite and the effect of the external edge can be neglected. The general solutions for stress components in Paper V follow those in the works of Lekhnitskii [54] and Zhang and Ueng [55].

A Cartesian coordinate system $x$-$y$ is set up, as shown in Figure 7, such that the displacement of the pin is in the direction of the positive $x$ axis. The radii of the pin and the pin hole are the same and denoted by $r$. The resultant force $P$ from the pin on the plate is known. The principal material directions $x_0$ and $y_0$ of the plate are such that the angle between the direction $x_0$ and the coordinate axis $x$ is $\phi$, seen in Figure 7. The elastic moduli along the principal material directions $x_0$ and $y_0$ are $E_{x0}$ and $E_{y0}$, respectively. The shear modulus is $G_{x0y0}$ and the Poisson’s ratio is $\nu_{x0y0}$. Lekhnitskii derived the expressions for plane stresses by solving the governing biharmonic differential equation in the complex domain [54]. He obtained

\[
\sigma_x = 2\text{Re} \left[ \mu_1^2 \phi_1'(z_1) + \mu_2^2 \phi_2'(z_2) \right],
\]

\[
\sigma_y = 2\text{Re} \left[ \phi_1'(z_1) + \phi_2'(z_2) \right],
\]

\[
\tau_{xy} = -2\text{Re} \left[ \mu_1 \phi_1'(z_1) + \mu_2 \phi_2'(z_2) \right],
\]

where $\phi_k$ are complex stress functions, and the first order derivatives of these functions are with respect to $z_k = x + \mu_k y$. The quantities $\mu_k$ are the complex roots of the characteristic equation in the coordinate system $x$-$y$, while $\mu_{k0}$ are those in the principal coordinate system $x_0$-$y_0$. If the resultant force components are known, then according to Lekhnitskii [41] the displacements at the plate hole are given by

\[
u^* = \alpha_0 + \sum_{m=1}^{\infty} \left( \alpha_m e^{im\theta} + \bar{\alpha}_m e^{-im\theta} \right),
\]

\[
\beta_0 + \sum_{m=1}^{\infty} \left( \beta_m e^{im\theta} + \bar{\beta}_m e^{-im\theta} \right),
\]
where $i$ is the imaginary unit and $\theta$ is the angle with the positive $x$-direction, as shown in Figure 7. The general solution for the stress functions can be expressed as

\begin{align*}
\phi_1(z_1) &= A \ln \xi_1 + \frac{2(\bar{a}_1 q_2 - \bar{\beta}_1 p_2) + \omega r (i q_2 + p_2)}{2D \xi_1} + \sum_{m=2}^{\infty} \frac{\bar{a}_m q_2 - \bar{\beta}_m p_2}{D \xi_1^m}, \\
\phi_2(z_2) &= B \ln \xi_2 - \frac{2(\bar{a}_1 q_1 - \bar{\beta}_1 p_1) + \omega r (i q_1 + p_1)}{2D \xi_2} - \sum_{m=2}^{\infty} \frac{\bar{a}_m q_1 - \bar{\beta}_m p_1}{D \xi_2^m},
\end{align*}

where the bar symbol $'\bar{\cdot}'$ denotes the complex conjugate, $\xi_1$ and $\xi_2$ are functions of the complex variables $z_1$ and $z_2$, respectively, and $A, B, D, p_1, q_1, p_2$ and $q_2$ are constants determined by the material properties. Since only derivatives of $\phi_1(z_1)$ and $\phi_2(z_2)$ are of interest for the determination of stresses, the integration constants are neglected. All aforementioned parameters are defined in [45, 46].
4. Tubular implants

4.1 Coupled helical coils

4.1.1 Introduction

Major congenital defects of newborns, such as tracheoesophageal fistula, intestinal atresia, and abnormalities of the anus and rectum, are relatively common. They need to be treated by replacing abnormal segments of tubular organs with patient-specific implants with suitable properties. Coupled helical coils, which can provide tissue-like compliance while preventing the collapse of tubular structures, have the potential to serve as tubular implants.

In Paper I, we manufactured coupled helical coils using a resorbable biomaterial. For studying the coupled helical coils with sparse mesh, analytical method is not applicable, and one must resort to a numerical approach. We also developed a FE model for such structures with the goal of studying structural response under small load conditions. Tensile and three-point bending tests were carried out to verify the suitability of the model. Small irreversible strains were considered, and the sensitivity of the response of the structure to the stress-strain relation of the material was small. In the next section, coupled helical coils with denser and periodic mesh were analyzed, and analytical approach, which is always favorable, was employed.

4.1.2 Methods

Coupled helical coils were manufactured using a computer numerical control (CNC) lathe (Siemens) in combination with a filament winding machine (Waltritsch & Wachter) shown in Figure 8(a). The preformed polycaprolactone (PCL) filaments were wound onto a polytetrafluoroethylene-coated needle tube mandrel. To obtain a coupled helix structure, different winding directions were used in wrapping. The right-handed wound filaments had a pitch (i.e. the width of one complete turn, measured along the axis of the helix) which could be the same as or different from the left-handed wound pitch. See Figure 8(b). After the winding steps, the tubes were rotated in an oven at 72 °C during 2.5 minutes to facilitate the fusion of filaments at the intersections.

Coupled helical coils, an example of which is shown in Figure 9(a), were modeled by use of in-house software. The diameter of the PCL filaments
was 0.3 mm, while the average diameter of the helical coils was 4.3 mm. The pitch lengths were 1.0 and 2.0 mm, and the total length was 36.0 mm. These dimensions are the same as those of the manufactured specimens used in the experimental testing.

Figure 8. (a) Manufacturing setup and (b) coupled helical coils made by filament winding of melt-extruded PCL, with an inner diameter of 4.0 mm and tube wall thickness of 0.3 mm.

Ten-node quadratic tetrahedral elements were chosen in the validated in-house FE code. About five elements along the diameter of the circular cross-section were used, which was found to be sufficient for dealing with the induced moments in the tetrahedral elements. For all numerical simulations, a Cartesian coordinate system was specified with one coordinate axis coinciding with the axis of coupled helical coils. In the analysis of tensile properties, the following boundary conditions were imposed on the FE model:

Figure 9. (a) Geometrical model of coupled helical coils and (b) FE model for three-point bending of coupled helical coils.
• At one end of the coupled helical coils the translational degree of freedom in the axial direction and two rotational degrees of freedom in the directions of the other two coordinate axes were fixed.
• The other end was subjected to a prescribed displacement in the axial direction of the structure.

Similarly, the following boundary conditions were applied for modeling three-point bending. See Figure 9(b):
• All translational and rotational degrees of freedom of the two supports were constrained.
• Prescribed displacement perpendicular to the axis of coupled helical coils was applied to the central rigid support to deflect the sample.
• Contact in the tangential direction was frictionless, and rigid contact was prescribed in the normal direction.

FE models for coupled helical coils with different dimensions and material properties were also simulated to parametrically study the influence of different design parameters on the structural properties of coupled helical coils. Totally 36 groups of simulations (tension and three-point bending) were implemented to 18 coupled helical coils with different combinations of design parameters.

4.1.3 Results and discussion

Figure 10 shows the structural responses of coupled helical coils from experiments. It is observed that the trends of the load-displacement relations are linear, Figure 10(a) and (b), but there is a systematic scatter in the original load data for both tests. This scatter could be explained by the lack of resolution of the 10-N load cell at these low levels of applied loads. The stiffness values were determined by linear fitting of the linear parts of the load-displacement data. Four samples of the same size were tested in axial tension and in three-point bending. The average results were found to be 0.70 N/m (range of values within ±5%) and 0.99 N/m (range within ±6%) for tensile and bending stiffness, respectively. Comparisons of the structural responses of the coupled helical coils are shown in Figure 10(c). The tensile and bending stiffness values from the FE simulations are 0.63 and 0.93 N/m, respectively. The deviations are 10% and 6.1% for the tensile and the bending stiffness, respectively. Given the variability in experimental parameters and specimen dimensions, this is considered to be acceptable.

The differences between predicted and experimental values can be attributed to concave menisci and thickening which can be observed at the obtuse angle where the coil filaments are fused together. In the FE model, the helical coils are just overlapped at the intersections, so the volumes of
the intersections are smaller than for the real samples. The flattening of the circular filament cross-sections leads to an increase in second moment of area and bending stiffness in the deformation direction of the free beam segments between the intersections.

![Graphs and Bar Chart](image)

Figure 10. Load-displacement relation of coupled helical coils under (a) uniaxial tension and (b) three-point bending, and (c) stiffness values for mechanical tests and FE simulations of coupled helical coils. The error bars indicate the range from the smallest to the largest experimental value.

The influence of the geometrical parameters of coupled helical coils on the structural properties was studied. Results of structural simulations for coupled helical coils with different average coil diameters are shown in Figure 11(a). Three kinds of samples were used in the numerical simulations with average coil diameters 2.3, 3.3 and 4.3 mm, filament diameter 0.30 mm, and pitches 1.0 and 2.0 mm. The span between the outer supports was 18.0 mm. By comparing the stiffness of different samples with same material property (elastic modulus of the PCL filament), it was found that increasing average coil diameter of coupled helical coils results in a decrease of tensile and bending stiffness. It is also observed that structural stiffness decreases non-linearly with the coil diameter.
The influence of pitch of coupled helical coils was also analyzed. Tensile and bending stiffness of coupled helical coils versus elastic modulus for different pitch combinations are presented in Figure 11(b). The fixed pitch of one helical coil is 1.0 mm, and the varying pitch of the other coil is 2.0, 4.0 and 6.0 mm. The filament diameter of 0.3 mm and average coil diameter of 4.3 mm are kept constant. The relations between structural stiffness and pitch are found to be non-monotonic. Provided that the material property is kept constant, the tensile stiffness of coupled helical coils decreases when the varying pitch increases from 2.0 to 4.0 mm, but increases when the variable pitch goes from 4.0 to 6.0 mm. For bending stiffness, a similar variation with pitch as for the tensile stiffness is observed (dashed line in Figure 11(b)). This non-monotonicity may qualitatively be explained by a combination of stiffening mechanisms with opposing trends with respect to pitch: (i) higher density of intersections for one of pitches being smaller, and (ii) stretching rather than bending of filament segments for higher pitch values.

![Figure 11](image)

*Figure 11. Influence of (a) average coil diameter and (b) pitch of coupled helical coils on their tensile and bending stiffness.*
4.2 Coupled helical coils with reduced sparsity

4.2.1 Introduction

Coupled helical coils with much more than two opposite-wound helices, are also of great importance in medical applications. One of such structures is mesh tubes, e.g. the Pipeline Neuroendovascular Device shown in Figure 12, with relatively simple geometries. They possess outstanding performances like high stiffness-to-weight and high strength-to-weight ratios, e.g. [47, 48].

With the development of efficient manufacturing technologies, mesh tubes are becoming widely used in a variety of applications, such as geotextiles, sieves, and medical implants [49, 50]. For better understanding of the mechanical properties of mesh tubes, evaluation of the load-displacement behaviour for the overall structures is of great importance. Extensive investigations have been devoted to the mechanical properties of mesh tubes [60–62]. Most of such studies are based on experiments or numerical simulations. To the knowledge of author, there are few analytical studies of the mechanical properties of mesh tubes. No closed-form solutions seem to exist, even though it is often advantageous to formulate an analytical solution which is more efficient and straightforward for prediction of structural properties compared with numerical methods. Coupled helical coils of very sparse mesh were not amenable to analytical modeling, so one must resort to a FE approach, which is the work that has been carried out in Paper I. In Papers II and III, it deems to be suitable to develop simple closed-form solutions for assessing the stiffness properties of mesh tubes.

Figure 12. Pipeline® embolization device [63].

4.2.2 Modeling

**Computational homogenization**

The curved mesh may be simplified as a planar material, as shown in Figure 13(a), provided that the RVE size is small compared to the diameter of the tube. For mesh tubes made by opposite-wound helical coils, the geometry of the RVE is then characterized as a rhombus, shown in Figure 13(b), and quantified by the strut length, \( l \), and the intersection angle between struts, \( \phi \).
The cross section of beams is assumed to be circular with diameter \( d \). This is the case, e.g., for filaments in melt drawn thermoplastic meshes.

\[ \frac{d}{l} \]

(a)                                             (b)

*Figure 13.* Schematic illustrations of (a) the periodic mesh structure, and (b) the geometry of the RVE. The brown solid lines delineate the mesh structure, and the dashed black lines RVE boundaries.

For a periodic mesh structure, the number of unit cells in the RVE, characterized by the number of joints included, does not influence the effective elastic properties. Thus, we only analyzed an RVE including one joint, as depicted in Figure 13(b). With computational homogenization, we obtained analytical equations for all effective elastic properties in *Paper II*,

\[
E_1 = \frac{3\pi(d/l)^3E \sin(\varphi/2)}{128 \cos^3(\varphi/2) + 6\pi(d/l)^2 \cos(\varphi/2) \sin^2(\varphi/2)},
\]

\[
E_2 = \frac{3\pi(d/l)^3E \cos(\varphi/2)}{128 \sin^3(\varphi/2) + 6\pi(d/l)^2 \sin(\varphi/2) \cos^2(\varphi/2)},
\]

\[
\nu_{12} = \frac{16 + 3(d/l)^2}{16 \cot^2(\varphi/2) + 3(d/l)^2},
\]

\[
G_{12} = \frac{\pi E(d/l)(16 \sin^2 \varphi + 3(d/l)^2 \cos^2 \varphi)}{256 \sin \varphi},
\]

where \( d/l \) is the strut diameter-to-length ratio (inverse aspect ratio).

**Analytical model**

Subsequent to computational homogenization, the mesh tube was converted to a thin-walled orthotropic tube with equivalent material properties as the original structure. Since the wall thickness was negligibly small compared with the tube diameter, a plane state of stress was assumed. A small section of the tube was simplified to a thin plate and studied. For the plate, the in-
plane constitutive equation is expressed in terms of the compliance matrix,

\[
\begin{bmatrix}
\varepsilon_1 \\
\varepsilon_2 \\
\gamma_{12}
\end{bmatrix} = \begin{bmatrix}
\frac{1}{E_1} & -\frac{\nu_{21}}{E_2} & 0 \\
\nu_{12} & \frac{1}{E_2} & 1 \\
0 & 0 & \frac{1}{G_{12}}
\end{bmatrix} \begin{bmatrix}
\sigma_1 \\
\sigma_2 \\
\tau_{12}
\end{bmatrix},
\] (10)

where 1 and 2 are indices for the in-plane principal material directions. Direction 1 coincides with tube axis. The elastic parameters \(E_1, E_2, \nu_{12}, \nu_{21}\) and \(G_{12}\) take the effective values expressed in Eq. (9).

The symmetries of structure indicate that one of the material directions of the equivalent continuum tube is parallel to its axial direction. Given that the lateral contraction of the small section is free, the corresponding strain can be expressed simply as \(\varepsilon_1 = \sigma_1/E_1\) under axial loading. By integrating \(\sigma_1\) over the tube cross section, one can determine the stiffness relation between the applied axial force and the resulting deformation. Replacing \(E_1\) with the corresponding effective elastic modulus acquired from computational homogenization, one can reformulate the axial stiffness as

\[
k_t = \frac{3En^3d^4H\sqrt{H^2 + 4\pi^2R^2}}{L[3H^2n^2d^2 + 64\pi R^2(H^2 + 4\pi R^2)]}.
\] (11)

By use of anisotropic beam theory, we can obtain the effective bending stiffness of the mesh tube as well. Since the primary material direction is parallel to the tube axis, the effective bending stiffness is solely dependent on the axial elastic modulus. Substituting the expression of the effective elastic modulus \(E_1\), we obtain the bending stiffness as

\[
\langle EI \rangle = \frac{6ER^2n^3d^4H\sqrt{H^2 + 4\pi^2R^2}}{3\pi H^2n^2d^2 + 64\pi^2 R^2(H^2 + 4\pi^2R^2)}.
\] (12)

4.2.3 Results and discussion

The filament material of the tubular structures was assumed to be linear elastic with an elastic modulus of 377 MPa and a Poisson’s ratio of 0.30, which can represent PCL filaments in a biomedical implant [64]. The geometry of the tube structure is defined by two sets of helical coils with the same geometrical parameters but different helical orientations, i.e. z-wound and s-wound.
Mesh sparsity

Mesh sparsity, denoted as $l/R$, was interpreted as the size of the RVE relative to the tube structure. Structural stiffness was calculated with the analytical model for mesh tubes with $l/R$ varying from 0.1 to 0.87. For FE simulations, the stiffness properties were modeled for four discrete values of the mesh sparsity: 0.12, 0.18, 0.24 and 0.36. The FE models were created by taking the total number of helical coils forming the mesh tubes $n$ as 36, 24, 18 and 12, respectively [64].

From Figure 14, an expected decrease of the axial and bending stiffness of mesh tubes as $l/R$ increases is observed. This is due to the increase of mesh sparsity of the structure. For a sparse mesh tube (high $l/R$), the stiffness is obviously lower than for a tube with denser structure.

![Graph](image)

*Figure 14. Stiffness versus mesh sparsity relations predicted by the analytical method and FE simulations for (a) axial loading, and (b) bending.*

The difference between the analytical and numerical results increases with increase of the ratio $l/R$, i.e. for sparser meshes. This is because with
the increase of $l/R$, the relative size of RVE is becoming larger and eventually comparable in size with the tube radius. The homogenization procedure in the two-stage model is not valid any more, since the length scales are too close [35]. This comparison implies that for use of the analytical model, the mesh sparsity must be within a suitable window. In the present case, the difference between the results from the analytical and numerical models is smaller than 10% for $l/R$ below approximately 0.2. A lower boundary for $l/R$ should be controlled by the applicability of the beam theory, i.e. strut slenderness, in the mesh homogenization.

**Coil slenderness**

The coil slenderness is defined as diameter-to-length ratio $d/l$ of the beam between two adjacent intersections. The stiffness properties were calculated from the analytical model for mesh tubes with $d/l$ varying from 0.01 to 0.27. For FE simulations, geometrical models were generated for mesh tubes with a number of coils of 36. The coil diameters were taken as 0.01 mm, 0.014 mm, 0.028 mm and 0.04 mm, which corresponds to $d/l$ of 0.06, 0.08, 0.15 and 0.22, respectively. From Figure 15(a) and (b), it is found that the axial and bending stiffness of mesh tubes increase when the ratio $d/l$ increases. With the other dimensional parameters constant, a higher ratio $d/l$ implies mesh tubes with thicker beams between intersections. Thus, the stiffness of the mesh increases, which contributes directly to the axial and bending stiffness of the tube. The analytical results deviate more from the numerical results with increasing ratio $d/l$. This is because in the homogenization procedure, the beam assumption for helical coils is violated if the aspect ratio of struts is too small. Therefore, the coil slenderness should be sufficiently large for use of the analytical model. To illustrate this, a comparison of computational homogenization results is shown between RVEs of beam elements and solid elements. The homogenization results for the elastic modulus $E_1$ is taken as an example and shown in Figure 16.

The intersection angle was fixed to be 60°. The diameter of struts in the RVE was kept as 0.3 mm, while the strut length varied to obtain a diameter to length ratio $d/l$ from 0.03 to 0.3. For homogenization using solid element RVEs, effective elastic properties were calculated for six discrete ratios, i.e. 0.03, 0.04, 0.06, 0.12, 0.15 and 0.24. The difference between beam and solid element results is found to increase monotonically with increasing values of $d/l$, Figure 16. The smallest and largest differences are about 4% and 28%, respectively. Thus, homogenization using beam elements and the bending model [65] becomes increasingly inaccurate as the ratio $d/l$ increases to large values. This is because when $d/l$ increases, i.e. strut length decreases, the joint is not negligible in size compared with the inter-joint struts. Significant elastic energy is stored by strains in the joint volume and not only in the deformation of the struts between the intersections. Thus, beam elements do not connect point-like joints in the RVE. To account for the shear stresses in
the beam when the diameter-to-length ratio is large, computational homogenizations with Timoshenko beam elements were also carried out. The accuracy of the results was not improved so much compared with Euler-Bernoulli beam element RVEs. In this case, the more complex deformation pattern and triaxial state of stress in the joint must be analyzed numerically using solid element RVEs to obtain accurate predictions of the elastic properties of the mesh.

![Graphs](a) and (b)

**Figure 15.** Stiffness versus coil slenderness relations predicted by the analytical method and FE simulations for (a) axial loading, and (b) bending.

**Intersection angle**

In this section, mesh sparsity and slenderness were kept approximately constant. Since the number of coils is an integer and the geometrical parameters are mutually dependent, only a discrete number of values of intersection angle were used in the simulations and the remaining geometrical parameters were adjusted to keep the ratios $d/l$ and $l/R$ approximately consistent as shown in Table 1. The variable geometrical parameters of the four different mesh tubes are listed in Table 2.
Figure 16. Influence of the beam diameter to length ratio $d/l$ on the effective elastic modulus $E_t$ predicted by the three methods with a fixed intersection angle of $60^\circ$.

Table 1. Geometrical parameters used in the analysis.

<table>
<thead>
<tr>
<th>Section</th>
<th>Mesh sparsity</th>
<th>Coil Slenderness</th>
<th>Intersection angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>$l/R$</td>
<td>Variable</td>
<td>Constant</td>
<td>Constant</td>
</tr>
<tr>
<td>$d/l$</td>
<td>Constant</td>
<td>Variable</td>
<td>Constant</td>
</tr>
<tr>
<td>$\varphi$</td>
<td>Constant</td>
<td>Constant</td>
<td>Variable</td>
</tr>
<tr>
<td>$L$ [mm]</td>
<td>24.0</td>
<td>24.0</td>
<td>24.0</td>
</tr>
<tr>
<td>$R$ [mm]</td>
<td>1.5</td>
<td>1.5</td>
<td>-</td>
</tr>
<tr>
<td>$H$ [mm]</td>
<td>9.0</td>
<td>9.0</td>
<td>9.0</td>
</tr>
<tr>
<td>$d$ [mm]</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$n$</td>
<td>-</td>
<td>36</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 2. Geometrical parameters of mesh tubes with different intersection angles.

<table>
<thead>
<tr>
<th>$R$ [mm]</th>
<th>$d$ [mm]</th>
<th>$n$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5</td>
<td>0.0140</td>
<td>36</td>
</tr>
<tr>
<td>2.0</td>
<td>0.0188</td>
<td>32</td>
</tr>
<tr>
<td>3.0</td>
<td>0.0280</td>
<td>30</td>
</tr>
<tr>
<td>5.0</td>
<td>0.0467</td>
<td>24</td>
</tr>
</tbody>
</table>

The axial and bending stiffness results for the mesh tubes are presented in Figure 17(a) and (b), respectively. Orange stars show the numerical results while the orange squares show the analytical results. From the curves, it is found that the relative differences between two methods with respect to numerical results is becoming smaller when the intersection angle $\varphi$ increases from $32^\circ$ to around $90^\circ$ for both axial and bending stiffness. The changing trend of stiffness properties indicates the influence of relative intersection volumes. When the intersection angle is $90^\circ$, i.e. the two coils cross each other perpendicularly, the intersection volume is at its minimum compared with the volume of the rest of the beams between the intersections. Thus, the
triaxial stresses in intersections can be ignored. With decrease of $\varphi$, the relative intersectional volume becomes non-negligible. The stress state is triaxial in the intersectional volume and generally not amenable to analytical treatment. Only if the intersection volume is relatively small, the beam element RVE is representative.

Figure 17. Stiffness-intersection angle relations predicted by the analytical method and FE simulations (a) axial stiffness (b) bending stiffness.

To further verify the aforementioned finding, the influence of the intersection angle, on microscopic level, on the homogenization results is also illustrated. The results are illustrated in Figure 18. The reference and deformed configurations of the beam element RVE under a uniform macroscopic stress in horizontal direction are also shown in Figure 18(a) for three different intersection angles (from left to right $\varphi = 60^\circ$, $90^\circ$ and $120^\circ$). The reference state is indicated by dotted black lines, and the deformed state is shown by solid red lines. The deformed RVE shows that all the four struts
were bent under a uniform strain field in the horizontal direction. The larger intersection angle, the more the struts were deflected in bending, which leads to a reduction in the effective elastic modulus.

![Graph](image.png)

Figure 18. Comparison between predicted results from computational homogenization and the bending model as a function the intersection angle of the RVE with a fixed beam diameter-to-length ratio \(d/l\) of 0.12 for the effective elastic properties: (a) the elastic modulus \(E_1\) and (b) the shear modulus \(G_{12}\).

For both elastic modulus \(E_1\) and shear modulus \(G_{12}\), the relative difference reaches a minimum value at \(\varphi = 90^\circ\), and increases when \(\varphi\) becomes smaller or larger than this value, Figure 18(a) and (b). The comparisons between the analytical results (from the bending model and the homogeniza-
tion using beam elements) and the solid element results indicates that for small/large intersection angles (close to 0° or 180°, respectively), the predictions of the analytical models are inaccurate due to an inadequate description of the joint deformation. This can be explained by the underlying assumptions of beam theory and the size of the joint volume in the RVE. For small or large intersection angles, the overlapping volume of the joint becomes relatively large, and the effective free strut length between the joints decreases. With a larger relative joint volume, the contribution of the beam deformation to the overall mesh deformation diminishes.
5. Mandible implant

5.1 Investigation of mandible implant failure

5.1.1 Introduction
Congenital malformations of mandibles and defects after traumatic injuries or tumor resection frequently result in functional and aesthetic problems. The method of choice today for reconstructing extensive mandibular defects uses a free vascularized fibula flap [66,67]. It includes considerable morbidity for the patient and is a time consuming and costly procedure which can only be performed by surgeons with specific competence. The use of titanium scaffolds for patients who do not require soft tissue transfer has the potential to substantially reduce the drawbacks of a micro-vascular reconstruction [66,68]. However, to the best of our knowledge, the design of such implants nowadays is much dependent on the experience of surgeons [69]. One way to prevent failures is to employ preoperative FE analysis to predict structural strength and assist the design of implants [70–72]. In Paper IV, we investigated through FE simulations the fracture of a titanium mandibular implant from a clinical case.

5.1.2 Methods
The preoperative CT data were treated using metal deletion technique code to remove metallic artifacts involved in certain CT slices. It uses a linear interpolation algorithm, forward projection and an edge-preserving blur filter to eliminate noise and streaking artefacts [50]. After preprocessing, the mandible bone regions were extracted using a global threshold-based segmentation method. The porous trabecular bone was left as a cavity, since its relatively low stiffness has negligible effect on the stress distribution in the implant [73–75]. Subsequently, 3D mandible objects were generated by a region-growing method and smoothed for meshing.

The mandible implant consisted of two distinct but integrated parts: the load-bearing plate and the scaffold. The thickness of the load-bearing plate was 1.6 mm and the total length was about 8 mm. To avoid the computational difficulties caused by direct meshing of the scaffold, it was converted into a homogenized continuum through computational homogenization, with a RVE as shown in Figure 19. The RVE comprises four titanium alloy struts.
with length 10.4 mm and diameter 0.70 mm which follow the space diagonals of a cube and intersect in the center as shown in Figure 19. The struts were made of a titanium alloy with Elastic modulus 110 GPa and Poisson’s ratio 0.35 [72].

Figure 19. FE model of the RVE of the scaffold (6.0 mm × 6.0 mm × 6.0 mm).

5.1.3 Results and discussion
The von Mises stress field in the load-bearing plate is shown for the isotropic cortical bone model in Figure 20(a) and for the orthotropic cortical bone model in Figure 20 (b) under the rightmost distal molar biting. Warm colors stand for high stress levels and cold colors for low stress levels. The maximum stresses are 161 MPa and 141 MPa for the isotropic and orthotropic cortical bone, respectively. Both are located in the vicinity of the screw hole IV shown in Figure 21(b). The elastic properties of the bone indeed have a non-negligible effect on the magnitude of stress concentration, but not on the location for the present geometry.

Figure 20. Calculated von Mises stress [MPa] on the load-bearing plate for (a) isotropic and (b) orthotropic cortical bone material properties.
It can be seen in Figure 20 that the stress fields in the load-bearing plate are quite similar for the two models. Moreover, the displacement fields for the two models were found to be quite similar. Since the maximum difference in magnitude was limited to 14%, only the orthotropic elastic material model is employed later. It also represents the most realistic model, since cortical bone is known to be a highly anisotropic material [76].

With the discrete scaffold replaced by the homogenized scaffold, a FE model with all essential input parameters was used to simulate mandible mastication. Effective elastic material properties (Elastic moduli, Poisson’s ratios and shear moduli) in the directions of the three coordinate axes in Figure 19 are shown in Table 3. Different biting modes, suggested by downward arrows in Figure 21(a), were simulated to analyze the influence of biting site on the stress field.

![Figure 21](image)

(a)

(b)

**Figure 21.** (a) Different mastication modes for FE simulations and (b) numbering of screws holes in the load-bearing plate.

**Table 3. Orthotropic elastic properties of the homogenized scaffold calculated by computational homogenization.**

<table>
<thead>
<tr>
<th>Elastic moduli [MPa]</th>
<th>Poisson’s ratios</th>
<th>Shear moduli [GPa]</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_X$</td>
<td>$E_Y$</td>
<td>$E_Z$</td>
</tr>
<tr>
<td>$v_{XY}$</td>
<td>$v_{YZ}$</td>
<td>$v_{XZ}$</td>
</tr>
<tr>
<td>$G_{XY}$</td>
<td>$G_{YZ}$</td>
<td>$G_{XZ}$</td>
</tr>
<tr>
<td>Equivalent continuum</td>
<td>83.7</td>
<td>83.7</td>
</tr>
<tr>
<td></td>
<td>0.49</td>
<td>0.49</td>
</tr>
<tr>
<td></td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td></td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td></td>
<td>1.0</td>
<td>1.0</td>
</tr>
</tbody>
</table>

The von Mises stress fields in the load-bearing plate due to mastication are shown in Figure 22. For biting carried out with the leftmost distal tooth, Figure 22(a), the largest von Mises stresses occur near screw hole II, Figure 21(b), of the plate. When the biting location moves towards the implant, the largest von Mises stresses move towards screw hole IV, as seen in Figure
The maximum von Mises stress in the load-bearing plate is 89 MPa under biting with leftmost distal molar and increases to 129 MPa and 141 MPa under middle incisor biting and rightmost distal molar biting, respectively. Correspondingly, the maximum stress at screw hole IV rises from 51 MPa to 97 MPa and 141 MPa, respectively. Thus, significant stress concentration occurs at screw hole IV for rightmost distal molar biting.

![Figure 22](image)

Figure 22. Calculated von Mises stress [MPa] field in the load-bearing plate by FE simulations for (a) biting with the leftmost distal molar, (b) biting with the incisors, and (c) biting with the rightmost distal molar.

The fractured plate, of which the three-dimensional geometry was reconstructed from CT data after implant failure, is shown in Figure 23. The location of the implant failure and that of the largest von Mises stress predicted by the FE simulations in Figure 22(c) agree well. However, the largest von Mises stress 141 MPa is well below the yield strength 800 MPa of the titanium alloy used for the implant. This might be explained by the existence of manufacturing defects in the load-bearing plate which reduce its strength.
For example, the bonding of titanium powder particles in the plate manufactured by electron beam melting (EBM) might not be perfect throughout the plate, especially in regions with complex geometry [77–79].

Figure 23. Three-dimensional geometry of the fractured load-bearing plate.

5.2 Stress distributions around screw holes

5.2.1 Introduction

In Paper IV, the stress analysis on the whole implants was carried out through numerical simulations which gave an overall picture of the implant. We also needed to look into local biomechanical phenomena during the design of implants. One phenomenon is bone loss, which is observed in patients after fracture treatment, especially when load bearing plates are used or bone grafting is involved. Resorption of bone surrounding the screws will consequently result in screw loosening, which in turn causes instability, non-union of the fracture infection and occasionally osteomyelitis [80,81]. This leads to the need for surgical revision removing fixation material and providing new adequate fixation with plates and screws [82]. The cause of the failure is most often due to incorrect analyses by surgeons about the requirement of biomechanical stability in a specific fracture location and pattern and/or the wrong choice of material for osteosynthesis not providing mechanical stability. From mechanical perspective, bone loss is believed to be associated with high peak stresses at the interface [83,84]. Therefore, in Paper V, stress distributions around screw holes were studied in an analytical way for preventing implant failures.

5.2.2 Modeling

The human mandible consists of two layers of cortical bones and one medium layer of cancellous bone. In Paper V, the mandible is presented by a laminated plate comprising three layers. The region-based elastic properties for cortical bone, presented in Table 4, are taken from [85] which is based on the study of Schwartz-Dabney and Dechow [76]. The cancellous bone in the mandible is assumed to be elastically isotropic, with an elastic modulus of
1.37 GPa and a Poisson’s ratio of 0.3 [86]. The effective elastic properties of
the mandible are calculated through classical laminate theory (e.g. [87]), i.e.,

\[
E_{x0} = \frac{A_{11}A_{22} - A_{12}^2}{A_{22}H}, \quad E_{y0} = \frac{A_{11}A_{22} - A_{12}^2}{A_{11}H},
\]

\[G_{x0y0} = \frac{A_{66}}{H}, \quad v_{x0y0} = \frac{A_{12}}{A_{22}},\]

where \(H\) is the thickness of the bone; \(A_{ij}\) are components of the stiffness ma-
trix \(A\) defined as,

\[
A_{ij} = \sum_{k=1}^{3} (C_{ij})_k t_k,
\]

in which \((C_{ij})_k\) are the stiffness matrix components of the \(k\)th layer and \(t_k\) is
the thickness of the \(k\)th layer.

### Table 4. Orthotropic elastic properties of cortical bone [85].

<table>
<thead>
<tr>
<th>Positions in mandible</th>
<th>Elastic moduli [GPa]</th>
<th>Poisson’s ratios</th>
<th>Shear moduli [GPa]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(E_1)</td>
<td>(E_2)</td>
<td>(E_3)</td>
</tr>
<tr>
<td>Symphysis</td>
<td>20.49</td>
<td>16.35</td>
<td>12.09</td>
</tr>
<tr>
<td>Body</td>
<td>21.73</td>
<td>17.83</td>
<td>12.70</td>
</tr>
<tr>
<td>Angle</td>
<td>23.79</td>
<td>19.01</td>
<td>12.76</td>
</tr>
<tr>
<td>Ramus</td>
<td>24.61</td>
<td>18.36</td>
<td>12.97</td>
</tr>
<tr>
<td>Condyle</td>
<td>23.50</td>
<td>17.85</td>
<td>12.65</td>
</tr>
<tr>
<td>Coronoid</td>
<td>28.00</td>
<td>17.50</td>
<td>14.00</td>
</tr>
</tbody>
</table>

Stress distribution in an orthotropic plate with circular openings has been
well studied by Lekhnitskii using a complex stress function method. For the
case where the radii of pin and pin hole are identical and the resultant in-
plane force is known, the general solution of stress functions are given by
Eq. (8) if the displacements at the plate hole are expressed by Eq.(7). It is
assumed that the major displacement boundary conditions is formulated as,

\[
\begin{align*}
    u_1 &= \frac{u_0}{c_1}, \quad v = 0 \quad \text{when} \quad \theta = \frac{\pi}{2},
    \\
    u_2 &= \frac{u_0}{c_2}, \quad v = 0 \quad \text{when} \quad \theta = -\frac{\pi}{2},
    \\
    (u_0 - u) \cos \theta &= v \sin \theta \quad \text{when} \quad -\frac{\pi}{2} < \theta < \frac{\pi}{2},
\end{align*}
\]
where \( u \) and \( v \) are displacements of points on the plate hole circumference on the \( x \)- and \( y \)- axes. \( u_0, u_0/c_1 \) and \( u_0/c_2 \) are displacements of points \( M, M_1 \) and \( M_2 \), shown in Figure 7, respectively. Furthermore, with the friction coefficient denoted by \( f \), certain stress boundary conditions on the plate-hole circumference are also introduced [55],

\[
\tau_{r\theta} = 0 \quad \text{and} \quad \sigma_{rr} = 0, \quad \text{at} \quad \theta = \pm \frac{\pi}{2},
\]

\[
\int_{\frac{-\pi}{2}}^{0} \tau_{r\theta} r \, d\theta - \int_{\frac{\pi}{2}}^{0} \sigma_{rr} r \, d\theta = f \left[ \int_{\frac{-\pi}{2}}^{0} \tau_{rr} r \, d\theta + \int_{\frac{\pi}{2}}^{0} \sigma_{rr} r \, d\theta \right]. \tag{16}
\]

Substituting the general stress components into the boundary conditions in Eqs. (15) and (16), one can determine unknown parameters and then the stress components

\[
\sigma_{rr} = \frac{P_x}{\pi Sr} \sum_{k=1}^{3} \left( C_k^\sigma \cos(2k + 1)\theta + S_k^\sigma \sin(2k + 1)\theta \right),
\]

\[
\tau_{r\theta} = \frac{P_x}{\pi Sr} \sum_{k=1}^{3} \left( C_k^{\sigma} \cos(2k + 1)\theta + S_k^{\sigma} \sin(2k + 1)\theta \right), \tag{17}
\]

\[
\sigma_{\theta\theta} = \frac{P_x}{2r} (\sigma_{\theta 1} + \sigma_{\theta 2} + \sigma_{\theta 3} + \sigma_{\theta 4} + \sigma_{\theta 5}),
\]

where the coefficients \( C_k^\sigma, S_k^\sigma, C_k^{\sigma}, S_k^{\sigma} \) and \( \sigma_{\theta k} \) are determined and given in the Appendix of Paper V.

5.2.3 Results and discussion

The stress distributions around the screw hole depend on the elastic properties of the cortical and the cancellous bones, the friction coefficient, and the loading direction. To illustrate the influence of different parameters, we have determined numerical values of the stresses for the different regions of mandibular bones based on the analytical model. The friction coefficient between bones and porous titanium alloy is around 0.6 [88]. Thus, the friction coefficient \( f \) in present work is taken as 0, 0.2, 0.4, 0.6 and 1, where the frictionless case corresponds to an unbonded screw. With the increase of the friction coefficient \( f \), the screw is considered to become increasingly bonded. For each value of the friction coefficient, the profiles of the radial normal stress \( \sigma_{rr} \) and the circumferential shear stress \( \tau_{r\theta} \) are calculated for different cross angles \( \phi \), i.e. \( 0^\circ, 15^\circ, 30^\circ, 45^\circ, 60^\circ, 75^\circ \) and \( 90^\circ \) between the loading direction and the direction of the main material axis.
The relative thickness of the cortical bone layer is defined as,

\[ t_r = \frac{2t_{\text{cort}}}{2t_{\text{cort}} + t_{\text{can}}}, \]

where \( t_{\text{cort}} \) is the thickness of cortical bone layer and \( t_{\text{can}} \) is the thickness of the cancellous one. The relative thickness varies for different regions of the mandible bone. For simplicity, the relative thickness for each of the superficial cortical bone layers is assigned as 0.2, 0.4, 0.5, 0.6, 0.8 and 1.0 for symphysis, body, angle, ramus, condyle and coronoid, respectively [29]. For a relative thickness 1.0, the bone is comprised only of the stiff cortical bone, which can be assumed to be the case coronoid part of the mandible. Due to symmetry the mandible bone is considered to be orthotropic, and the calculated effective elastic properties in different regions are listed in Table 5.

Table 5. Effective orthotropic elastic properties of different regions of the mandible.

<table>
<thead>
<tr>
<th>Mandible</th>
<th>( E_{x_0} ) [GPa]</th>
<th>( E_{y_0} ) [GPa]</th>
<th>( G_{x_0y_0} ) [GPa]</th>
<th>( v_{x_0y_0} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Symphysis</td>
<td>5.20</td>
<td>4.37</td>
<td>1.80</td>
<td>0.33</td>
</tr>
<tr>
<td>Body</td>
<td>9.51</td>
<td>7.95</td>
<td>3.30</td>
<td>0.34</td>
</tr>
<tr>
<td>Angle</td>
<td>12.58</td>
<td>10.20</td>
<td>4.06</td>
<td>0.30</td>
</tr>
<tr>
<td>Ramus</td>
<td>15.31</td>
<td>11.57</td>
<td>4.44</td>
<td>0.28</td>
</tr>
<tr>
<td>Condyle</td>
<td>19.08</td>
<td>14.56</td>
<td>5.83</td>
<td>0.24</td>
</tr>
<tr>
<td>Coronoid</td>
<td>28.00</td>
<td>17.50</td>
<td>7.15</td>
<td>0.23</td>
</tr>
</tbody>
</table>

Average stresses over the mandible cross section

It is found that the differences between stress distributions in different regions are not significant for the chosen values of \( f \) and \( \varphi \). The angle region in the mandible, where 25% of fractures occur, has been chosen to study the influence of the friction coefficient \( f \) and the cross angle \( \varphi \) in more detail. Stress distributions are shown in Figure 24. Different colors represent different values of the friction coefficient \( f \) and different brightness intensities stand for different values of the cross angle \( \varphi \).

The stress distribution curves can be divided into approximately five groups with respect to friction for the normalized normal and shear stresses. With increasing friction coefficient, the level of the radial stress around the screw hole is gradually reduced. Simultaneously, the distribution for the normal stress in radial direction \( \sigma_{rr} \) changes with the high stress region spreading out bilaterally from \( \theta = 0^\circ \). Comparing the curves in each group, the effect of loading direction, \( \varphi \), can be observed under closer scrutiny. The maximum stress is varying only within a small range and the stress distribu-
tions remain relatively constant. The variation of the friction coefficient, \( f \), obviously has a larger impact on the stress state than the variation in cross angle, \( \varphi \) [89].

The critical point for possible damage, e.g. cracking, is at the location of maximum stress levels. To demonstrate the influence of friction and loading direction, the maximum stresses for angle region with respect to \( f \) and \( \varphi \) are compared. It is found that the differences of maximum stresses \( \sigma_{rr} \) and \( \tau_{r\theta} \) are much smaller when the angles are varying than when the friction coefficient is varying. Thus, not only for the stress distributions in Figure 24, but also for the maximum stresses, the friction plays a more important role than the direction of the load.

![Figure 24](image_url)

*Figure 24.* Distributions of normalized (a) normal and (b) shear stresses on a screw hole boundary in the angle region of a mandible. The friction coefficient takes values 0, 0.2, 0.4, 0.6 and 1, and the cross angle takes values 0°, 30°, 60° and 90°.
Potential applications of the analytical model

The current analytical method could also be used together with other methods to analyze bone remodeling or to predict failure. A comprehensive procedure is proposed by the flow chart in Figure 25. The analytical stress analysis is limited to the shadowed box. The schematic shows how this analysis could fit into a larger framework.

![Flow chart showing the procedure for predicting bone remodeling or failure from mandible implant screws.](image)

**Figure 25.** Schematic of a more comprehensive approach to prediction of bone remodeling or failure from mandible implant screws.
As is seen, the kinetics information of the mandible and the implant could firstly be obtained through a macro-level analysis using commercial software, e.g., see [90]. The in-plane loads applied to the screws are then used as input for the analytical stress analysis. Overall stress distributions around the screw hole are determined by superposing contributions from in-plane loads and other loads, e.g., moments and muscular forces. The stress analysis could be further used as a point of departure to investigate the bone remodeling and predict bone failure, through Wolff’s law and failure criteria, e.g., Tsai-Wu criterion, respectively. To keep in mind, this analytical model is mainly applicable for initial stage of bone remodeling and failure initiation, since the model is based on homogeneous properties of the bone.

For implementing failure criteria, it is preferable that the stress states in each bone layer are known, since failure criterions might be different for different bone structures. For example, the Tsai-Wu criterion is deemed to be more suitable than von Mises criteria for assessing the failure of cortical bones which are known as orthotropic materials [91]. The local stresses in a specific layer, e.g., the cortical bone layer, are determined and analyzed. If only in-plane deformation is considered, the average strain vector \( \varepsilon_{\text{lam}} \) of the laminated bone are expressed as,

\[
\varepsilon_{\text{lam}} = H A^{-1} \sigma_{\text{lam}},
\]

where \( \sigma_{\text{lam}} \) is the average stress vector of the laminated bone in Cartesian coordinates. Then the constitutive equation of the cortical bone is employed to calculate the local stresses in the cortical bone layer by,

\[
\sigma_{\text{cort}} = D_{\text{cort}} A^{-1} \sigma_{\text{lam}},
\]

in which \( D_{\text{cort}} \) is the stiffness matrix of the cortical bone material under plane stress state in the \( x-y \) coordinate system.

The stress profiles around the screw hole in the cortical bone layer are demonstrated in Figure 26. Only the distributions of the normalized stresses for \( f = 0 \) and \( \varphi = 0^\circ \) and \( f = 1 \) and \( \varphi = 90^\circ \), which are two extreme cases. Shear stress \( \tau_{r\theta} \) does not exist for the frictionless case, and thus is not demonstrated. As is seen from the diagram, the differences between stresses in different regions are more distinguishable than those for average stresses over the cross section. Applying the same in-plane force to the screw, the stress levels in the different regions of cortical bone are different. The largest maximum stress level is observed in the symphysis region, and decreases with screw location moving towards coronoid region. If the same stress-based failure criterion is applied for cortical bone, failure is more likely towards the symphysis region since the stresses are the higher there.
Figure 26. Stress distributions on the boundary of screw hole in the different regions of the mandible in the cortical bone layer. The friction coefficient and the cross angle are: (a) 0 and 0° and (b) and (c) 1 and 90°.
6. Conclusions and future work

6.1 Conclusions

This thesis deals with stiffness properties and stress distributions related to two types of medical implants, tubular and mandibular, in the context of biomechanics. The main conclusions of the thesis can be summarized as follows:

Coupled helical coils can be manufactured in a well-controlled and reproducible manner by winding commercial PCL filaments. The good agreement obtained between numerical FE and experimental results for such devices suggests that FE simulations can be used as a versatile tool for design of tubular implants. On the basis of simulations, filament diameters and pitches can be suggested for target structural responses. With a parametric code for creating the geometry of coupled coils with arbitrary pitches and diameters, and automatic meshing, such simulations of structural response are not very time consuming.

Computational homogenization is shown to be a procedure that makes it possible to obtain analytical solutions for effective elastic properties of 2D mesh structures. It is more general and straightforward, and requires fewer algebraic manipulations, than derivation of analytical expressions using a beam bending model. On the basis of homogenization results, an analytical model was developed for coupled helical coils consisting of multiple helices. The key geometrical parameters that control the applicability of the analytical model have been identified through comparison with corresponding results of FE simulations. The present analytical model is found to be valid when the RVE is small compared to the macroscopic dimensions of the mesh tube, the beams between the intersections in the mesh are slender, and the volumes of the intersections are small relative to the volume of the RVE.

Numerical studies of implant failure carried out show that it is feasible to quantitatively investigate failure of a mandibular implant using FE simulations. They also confirm the feasibility of using CT image analysis, CAD, and FE simulation to predict stress fields in complex patient-specific implants and to assist in the design of such implants. Since additive manufacturing allows for geometrical designs virtually without limitations, this design approach could be complemented with structural optimization schemes to improve the mechanical performance of the implants under given constraints.
Parametric analyses of design parameters, such as the friction coefficient and the loading direction, were performed for screw-loaded holes in the mandible. They were based on an analytical model which allows much faster analyses than FE simulations. This model may also be implemented in a larger framework as a point of departure for predicting bone remodeling on the basis of Wolff’s law or for predicting failure on the basis of a suitable criterion. However, one should keep in mind that the proposed analytical model is valid only for in-plane loading of flat parts of the homogeneous bone, where effects of finite dimensions, bending moments and out-of-plane loads can be neglected.

6.2 Future work

The studies in this thesis aim at facilitating clinical application of new types of medical implants from a mechanical perspective. Most of them are of preliminary nature. In the future, more complex and detailed studies should be carried out.

Stiffness analyses of coupled helical coils made from bio-resorbable materials such as PCL have been carried out both numerically and analytically. The coupled helical coils are intended to degrade in vivo to allow corresponding generation of new tissues. It is of critical importance to maintain the desired stiffness properties during this degradation. Therefore, as a next step, parameters and variables associated with PCL degradation, such as elastic modulus and degradation rate, respectively, should be taken into account in the analyses.

Numerical analyses based on FE simulations are found to be effective for assessing stress distributions and predicting mandibular implant failures. The elastic properties of mandible bones have influence on the maximum values of stresses. To make more accurate predictions, more realistic elastic properties of specific patients are needed. A potential approach is to make use of the relation between CT data intensity and material density to extract elastic properties of bone. Other important input parameters are the muscular forces. The numerical values of such forces under different conditions should be calculated using commercial musculoskeletal software.

Image analysis techniques and FE simulations have been used collectively in various biomechanical studies. An important feature of medical implants in the future will be customization, i.e., implants will be fully adapted to the specific requirements of each patient. This could be achieved by developing design tools that combine CT diagnosis, image analysis, CAD, AM and FE analysis in a manner that is user-friendly for surgeons and other clinical practitioners.

Stress distributions around screw-loaded holes in bones have been determined analytically. A potential application of such analytical solutions is
evaluation of bone remodeling in a local region by use of Wolff’s law. This implies that the bones will become heterogeneous. Accordingly, a suitable analytical model should, if possible, be developed to account for the material properties of heterogeneous bone.
7. Summary in Swedish


Rörformiga implantat med kopplad spiralstruktur, som potentiellt kan ersätta rörformade organ såsom luftstrupor och urinrö, tillverkades genom att linda tråd av biokompatibel och resorberbar polykapsulaktom (PCL). Rörimplantatens drag- och böjstyvhet bestämdes genom numeriska simulerings med finita element (FE) samt genom mekanisk dragprovning och trepunktsböjning. Dimensioneringsparametrar som spiraldiameter och stigning studerades, varvid god överensstämmelse erhölls mellan numeriska och ex-

förutsäga intial benåterväxt eller benbrott med hjälp av lämpligt spänningskriterium.
Acknowledgements

This thesis was written based on the research study supported partly by Chinese Scholarship Council and performed at the Department of Engineering Sciences at Uppsala University. It would not have been completed without the contributions from many wonderful individuals.

First of all, I would like to express my sincere gratitude to my main supervisor Professor Kristofer Gamstedt, for bringing to Uppsala and guiding me through the whole Ph.D study with immense knowledge and great patience. Thank you for always being positive and encouraging even when things went slower than expectation. Over the years, I have learned a lot from you, especially the way of thinking as an independent researcher. Further, I would like also to thank my co-supervisor Associate Professor Urmas Valdek for providing valuable suggestions on the research work.

I would also like to express my sincere thanks to Professor emeritus Bengt Lundberg. It was a wonderful time to share an office with you, which gave me the chance to discuss and share varieties of things with you and learn from your enthusiasm and brilliant way of thinking. Thank you for not only providing constructive comments on research work, but also helping with other countless matters.

Many thanks to all the former and present colleagues in the Division of Applied Mechanics: Dr. Thomas Joffre, thank you for picking me up in Arlanda airport and sharing knowledge and resources on image analysis. Dr. Gabriella Josefsson, thank you for all the help since the beginning of my study in Uppsala. Dr. Nico van Dijk, thank you for all the support and important contribution in the homogenization manuscript. Fengzhen Sun, Shaohui Chen and Dan Wu, thank you for sharing lots of fun and interesting discussions. Former officemates Dr. Adrian Hernández Pérez and Associate Professor Ivón Hassel for all your help and encouragement. Peter Bergkvist, for all the help with the lab tests. Elias Garcia Urquia, for the fun moments and the wonderful trip to Barcelona.

I would like to thank all the coauthors and collaborators of the appended papers: Professor Jöns Hilborn, Dr. Ramiro Rojas, Jan Bohlin, Doctor Per Dérand, Associate Professor Lars-Erik Rännar, and Associate Professor Andreas Thor. Professor Emeritus Jan-Michaël Hirsch, thank you for coordinating the collaboration, and providing essential support from medical field.
Professor Cornelia Kober, thank you for your valuable suggestions and inspiration in CT image analysis.

Finally, my sincere gratitude is reserved to my whole family and relatives for your continuous love and support. My wife Yingying, thank you for always accompanying me no matter where I am. Your patience, support and encouragement have been utmost important for my Ph.D study.
References


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