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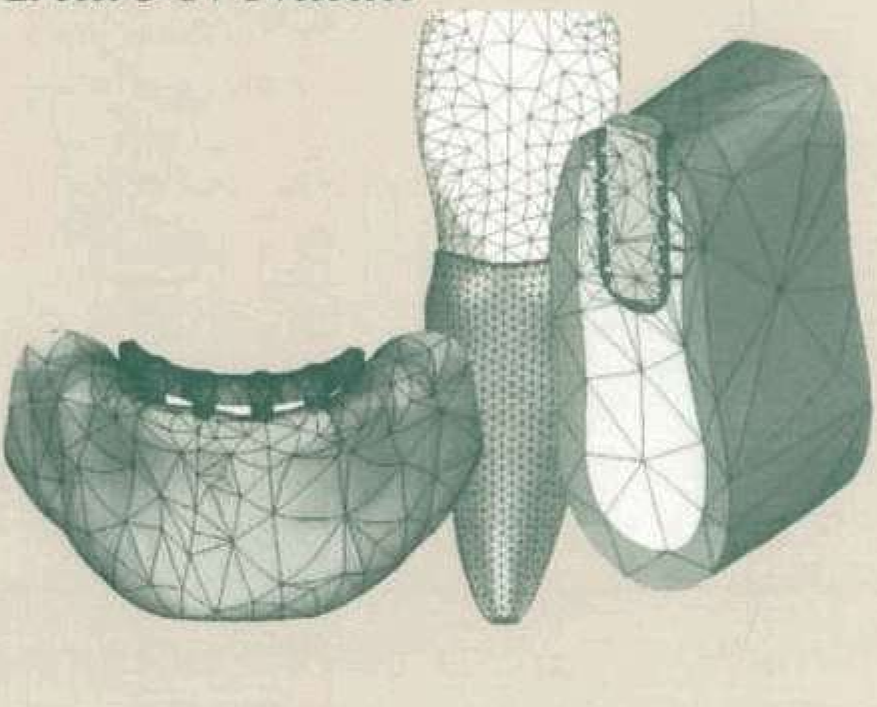

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Preface

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Thus my plan here is not to teach the method that everyone must follow in order to guide his reason, but merely to explain how I have tried to guide my own.

Those who set themselves up to instruct others must think they are better than those whom they instruct, and if they misguide them in the slightest they can be held responsible.

But, since I am proposing this work merely as a history or, if you prefer, a fable – in which, among a number of examples that may be imitated, there may also be many others where it would be reasonable not to follow them – I hope it will be useful for some readers without being harmful to others, and that everyone will be grateful for my frankness.

. . .

. . . I hope that those who use only their pure natural reason will be better judges of my views than those who trust only ancient books. For those who combine common sense and study – and I hope that they alone will be my judges . . .

. . .

I would say only that I decided to use the time that remains to me in life for nothing else except trying to acquire a knowledge of nature, from which one could draw some more reliable rules for medicine than those we have had up to now.

René Descartes, "Discourse on method"

I consider it essential to question my dedication to research and, once I am in the midst of it, to reflect on the outstanding privilege of treating the mechanics of biological tissues. I like to consider the approach to be taken on, aiming at the integration of all of the knowledge and competencies that are a part of the research. The significant complexity of biomechanical processes is the manifestation of a superior formulation. Nevertheless, problems that may at first appear insurmountable, can be successfully interpreted by means of an attentive and humble approach that can lead towards the definition of a realistic final configuration. The functional response of biological tissues is, in and of itself, a fundamental reference, which can then be used to access the mechanics within the biological phenomena being dealt with.

The strong desire to reach a solution, or the reduced potentiality of the resources adopted for the investigation, should not lead to inadmissible approximations. On the contrary, regardless of how they have been chosen, they must be evaluated for the implication they have on the reliability of the final result, and should represent a cautious passage towards a more complete interpretation. A comparison of the results deriving from subsequent models, whose accuracy has been improved by taking the characterising aspects into account, will tell us how appropriately the investigation has been carried out. These models will be milestones, which confirm that the right course has been taken.

With this in mind, the mechanics written in biological phenomena should be read leaving aside the fear of facing their enormous complexity; however, at the same time the researcher must also be guided by the precaution taught by experience. The researcher should not be tempted by immediate results, even if they are attractive, rather than seeking deeper insight into the subject at hand. To carry out research in this way, an ethical approach must be taken, coupling biology and mechanics by using the most updated methodology. Mechanics, physics and chemistry are strictly related to clinical practice for the evaluation of the operational reliability of the results obtained.

I intend to report part of the experience and results deriving from many years of activity, in research and education, regarding dental biomechanics. When presenting this work, I am faced with problems pertaining to form and depth with regards to different aspects of bioengineering, which must be treated while remaining compatible with clinical knowledge. The difference in the methods in these cultural areas makes it difficult to propose a unitary presentation of the problems dealt with. Nevertheless, great effort must be made to overcome this discrepancy, with the aim of arriving at a fruitful confrontation and moving towards a unitary definition.

The cooperation efforts between bioengineers and clinicians have proved to be a challenge. It is necessary to be realistic and consider the significant difficulties inherent in this situation. As Renè Descart stated, "If artisans cannot implement immediately the invention I explained, I do no think that, for that reason, it can be said to be defective. Since skill and practice are required to construct and adjust the machines that I described, even though no detail is omitted, I would be just as surprised if they succeeded on their first attempt as if someone were able to learn to play the lute very well in a single day, when they are provided with only a good tablature".

I hope that the final results of this challenge, rather than displease both engineers and clinicians, promote the substantial integration of interest and engagement in facing sophisticated biomechanical problems.

The structure of this work is based on the intention of describing a sequence of events that, in a general sense, should characterise the biomechanical analysis in the dental area. First of all, the mechanics of hard and soft biological tissues, namely the bone and periodontal ligament, is given. Following this characterisation of materials, the geometric configuration of the anatomical site is defined, using tomographic techniques, along with a description of pre-surgical procedures. A significant portion is devoted to the definition of the materials used in dental practice, with regard to both implantology and orthodontics, considering specific manufacturing techniques as well. In the same way, the clinical aspects are reported because of their relevance to practice in implantology and orthodontics. The numerical approach to the biomechanical analysis of dental problems is presented in order to describe the potentialities offered by numerical simulation. A summary of the mechanics of materials, in terms of basic formulation, is reported, as a fundamental reference for approaching the biomechanical aspects treated.

The outstanding complexity of biomechanical phenomena expresses a level of optimisation that seems inaccessible for our knowledge, and is source of wonder and respect. The careful consideration of the magnificence of this reality should move anyone involved in this investigation to humility, and to great dedication. Even if this involvement pertains to the definition a small portion of a problem, it could nonetheless represent a great achievement. To be aware of our own position within the field of knowledge constitutes a preliminary requirement for knowledge itself.

A discussion on method and knowledge becomes a unique task, passing through the ethics of the person, with the aim of achieving a common end. If my work could serve the purpose of a better integration of researchers and teachers who differ because of their scientific education, I hope it could also serve the purpose of helping create better understanding among the people themselves.

I would like to thank everyone that helped me to give substance to these thoughts. For this, I give my profession of gratitude.

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1 Mechanics of bone tissue

AN Natali, RT Hart, PG Pavan, I Knets

1.1 INTRODUCTION

The present chapter deals with the mechanics of hard tissues, namely cortical bone and trabecular bone. The chapter presents various aspects of experimental activities that have been developed for the investigation of the mechanical responses of bone tissue. Tissue properties depend on the environmental conditions of the tissue, including hydration, age of specimen, etc., as well as on mechanical loading conditions, such as the rate of loading, duration of loading, etc.

It is probably superfluous to underline that experimental testing in the field of biological tissues, in particular with regard to bone tissue, requires a sophisticated approach. In fact, the limited dimensions and acquisition procedures of vitro specimens make it difficult to interpret experimental results. Moreover, the mechanical characteristics of the tissue depend on many factors, such as temperature and moisture during testing, because bone specimens are subject to the degradation caused by environmental and biochemical conditions.

Experimental testing of bone is carried out in order to get a deeper knowledge of the mechanics of bone and also to create a database to be used to perform numerical analyses of biomechanical problems (Natali and Meroi, 1989).

Numerical methods, especially the finite element method, provide powerful tools for simulating and predicting the mechanical behaviour of biological tissues. They make it possible to obtain a detailed representation of many different factors that affect the biomechanical behaviour of bone: mechanical properties, shape, loading configuration, and boundary conditions. In order to control the validity of numerical techniques, numerical results and experimental data must be compared to ensure that the models represent the real behaviour of biological structures. The results obtained validated the finite element method as a fundamental approach to investigating phenomena such as implant bone interaction or even the remodelling of bone. These *in vivo* responses, including bone reaction to orthodontic procedures or to dental implant practice, are aspects of bone behaviour that are particularly pertinent to dental biomechanics.

The study of bone, a unique living structural material, requires an interdisciplinary approach to understand and quantify bone functions and adaptations. Bone's functional adaptation to mechanical loading implies the interpretation of a physiological control process. Essential components for this process include sensors for detecting mechanical response and transducers to convert these measurements to cellular response. The cellular response leads to gradual changes in bone shape and/or material properties, and once the structure has adapted, the feedback signal is diminished and further changes to shape and properties are stopped.

In order to make progress in understanding the complex response of living bone, the material and structural properties must first be quantified.

1.2 BONE

Bone is classified as a hard tissue, a definition that includes all calcified tissues. To adequately describe bone, a hierarchic scale must be identified in order to study its functional activity and properties. In fact, the macroscopic behaviour of bone is the reflection of a complex micro system relating to functions that determine the evolution of bone itself over time (Cowin, 1989). At the macroscopic level, namely for a large sample of material, bone strongly depends on the sample location and orientation as well as on specimen status and environmental factors.

At tissue level scale, bone has two distinct structures called cortical bone and trabecular bone. Cortical bone is the hard outer shell-like region of a bone and can further be classified as either primary or secondary. Primary cortical bone is made up of highly organized lamellar sheets, while secondary cortical bone is made up of sheets that are disrupted by the tunnelling of osteons centred around a Haversian canal. Trabecular bone, also called cancellous or spongy bone, is composed of calcified tissue, which forms a porous continuum.

Bone can be described as a complex of activities of three main types of cells: osteoblasts, osteoclasts and osteocytes. Osteoblasts are the cells responsible for new tissue production. Osteoclasts are related to the resorption of bone. Osteocytes are cells that are present in completely formed bone.

Mineralised microfibrils of collagen are recognised to be the components at an ultrastructural level. Their dimension is of the order of 3–5 μm. At a molecular level, three left handed helical peptide chains coiled into a triple helix form the tropocollagen molecule. These molecular structures have a dimension in the range of 1.5–280 nm (Katz, 1995)

Bone tissue is continuously renewed via a complex but well coordinated sequence of activities that results first in the replacement of primary bone by secondary (osteonal) bone tissue, followed by continual renewal of the secondary bone. Bone surfaces, including not only periosteal and endocortical surfaces, but also intracortical Haversian and trabecular surfaces, are the sites for cellular activity. The bone renewal process, called remodelling, depends on a vascular supply not only for oxygen and the exchange of nutrients and minerals, but also because the pre-osteoclasts, originating in the marrow, are present in the circulation before differentiating into active osteoclasts. The multi-nucleated osteoclasts adhere to bone surfaces with a characteristic ruffled border. This allows for the creation of a (permeable) sealed microenvironment where resorption occurs, bone mineral is dissolved and the collagen and other proteins are digested (Jee, 2001).

The local coupling of bone resorption followed by new bone formation during remodelling is not yet fully understood. However, it is known that bone renewal takes place as discrete packets of cortical or trabecular bone are destroyed and replaced by a group of osteoclasts and osteoblasts referred to as a BMU (Basic Multicellular Unit). As recently described by Jee (Jee, 2001), the BMU cycle includes six consecutive stages, resting, activation, resorption, reversal, formation, mineralization, that result in the coordinated removal of bone and the construction of a new structural bone unit, either an osteon or a trabecular packet (hemiosteon). Even if the key signals for each of these steps are not fully understood, the local mechanical environment and the local chemical environment (including hormones and growth factors) are both known to be important. In addition, since the renewal process is not perfect, only about 95 per cent of the removed bone is replaced (Jee, 2001), the bone structure becomes increasingly compromised over time.

Bone can also be stimulated to change its shape and size, a process called either modelling or net remodelling. Most modelling occurs during growth with changes in bone shape and size. However, even after maturity, bone may be stimulated by altering mechanical loading or by different agents that affects change of shape and/or material properties.

1.3 EXPERIMENTAL TESTING AND RESULTS

In spite of the fact that an official codification of experimental testing procedures has not been fully defined, the preparation of specimens follows an almost standard procedure. The main problem is to obtain *in vitro* specimens that should have, as far as possible, the same characteristics of the tissue *in vivo*, especially with regard to bone hydration (Ashman, 1989). Freezing specimens is probably the most common process used to maintain the original water content in bone. Furthermore, freezing has a marginal influence on bone mechanical properties.

The method of testing traditional engineering materials is also adopted for bone, paying attention to the small dimension of the specimens taken from *in vivo* bones (Reilly et al., 1974). Samples of the cortical portion of bone are usually about 5/5/15 mm, with square or circular cross section. The samples of cancellous bone have similar dimensions but are usually potted in acrylic at the ends.

Several tissue characteristics are highlighted in the following description of experimental testing results for cortical and trabecular bone.

1.3.1 Anisotropic characteristics of bone tissue

The anisotropic stiffness properties of cortical bone are revealed by simply loading the specimens in different directions. Values of the elastic parameters are found, depending on the loading direction, as reported in Table 1.1. In addition, the strength of bone depends on the loading direction and differs depending on compression or tension loads. Table 1.2 shows values which are usually assumed for yield and ultimate stress. They are obtained by applying axial loading and torsional loading to cortical bone specimens taken from human femur (Cowin et al., 1989). The maximum value of the yield stress is found for compression

Table 1.1 Average elastic constants for mandibular corpus in different zones

	Inferior	*Lingual*	*Buccal*
E_1 [GPa]	10.63	10.85	11.04
E_2 [GPa]	12.51	16.39	15.94
E_3 [GPa]	19.75	18.52	18.06
G_{12} [GPa]	3.89	4.59	4.31
G_{13} [GPa]	4.85	5.45	5.2
G_{23} [GPa]	5.84	6.49	6.45
ν_{12}	0.313	0.138	0.138
ν_{13}	0.246	0.338	0.322
ν_{23}	0.226	0.332	0.294
ν_{21}	0.368	0.178	0.257
ν_{31}	0.465	0.572	0.518
ν_{32}	0.356	0.357	0.326