Mechanical Properties of Biological Materials

M. Richetta and R. Montanari

Department of Industrial Engineering
University of Rome-Tor Vergata - Italy,
Mechanical Properties of Biological Materials

1. Growing need of new materials for:
   • Regenerative Medicine
   • Biomimetic Applications

2. BM properties becomes critical for BIOLOGICAL FUNCTIONS of MOLECULE, TISSUE, ORGANS

determines

RAPID EVOLUTION OF

BIOMATERIALS SCIENCE
Mechanical Properties of Biological Materials

ARTZ PENTAHEDRON

Schematic representation of:

a. contributing scientific fields

b. constraints/components in the study of biological systems
Mechanical properties of Biological Materials

Basic mechanical properties of materials:

**Strength** - the amount of stress required to permanently deform the material

**Stiffness** - the resistance to elastic deformations

**Toughness** - the ability to resist the propagation of cracks
Mechanical properties of Biological Materials

To characterize a BM we study its

STRUCTURAL PROPERTIES as

DEPENDENCE OF DEFORMATION ON APPLIED FORCES

This will be used to predict:
1) material properties in terms of stress and strain;
2) how “in vivo” tissue could behave.
BMs exhibit:

- **highly controlled structural features** (despite being composed of modest ingredients);

- **boast performances** (better than those of engineering materials);

- **adaptable compositions and structure**;

- **self repair and remodel**.
Mechanical properties of Biological Materials

Hard BMs are of less quality than the engineering ones but have better mechanical performance.

BMs are tough and strong (mutually exclusive properties in engineering materials), as HUMAN tooth is, since it is made of:
- an external “hard” layer: ENAMEL
- an internal “tough” one.
Dentin

Dentin exhibits **elastic**, **anelastic** and **plastic** behaviour.

The knowledge of its properties is essential in clinical dentistry:

- for understanding the effects of various restorative dental procedures;
- for predicting the effects of microstructure alterations due to caries, sclerosis and aging on tooth strength.
H-bonds between the polypeptide chains of collagen triple helix
Sample preparation

To study elastic/anelastic properties of dentine we studied it by Mechanical Spectroscopy and FIMEC indentation test.
Sample preparation

Human molars extracted from individuals (males 55-70 years old) as part of their dental treatment.

Disinfected by immersion in a solution of sodium hypochlorite in water for about 12 hours. Longitudinally sectioned in order to obtain 0.8 mm-thick slices. From these sections bar-shaped samples (length $L = 13 \div 16$ mm) have been cut for measurements.

Dentin density is different from point to point decreasing from the outer part to the inner one thus mechanical properties are not homogeneous.

Therefore, elastic modulus $E$ and $Q^{-1}$ obtained in present experiments represent average values.
Vickers Micro-hardness

Fimec Instrumented Indentation
Mechanical Spectroscopy

The $13\div16$ mm samples, mounted in free-clamped mode have been tested using the method of frequency modulation.
Damping factor $Q^{-1}$

\[ d = \frac{1}{k} \ln \frac{A_n}{A_{n+k}} \]

\[ Q^{-1} = \frac{d}{\pi} \]

Dynamic modulus $E$

\[ E = \frac{48\pi^2 P L^4}{m^4 h^2} f^2 \]

- $f$ resonance frequency
- $m$ constant ($m = 1.875$)
- $P$ material density
- $L, h$ length and thickness of the sample
Experiments above room temperature

Heating-cooling cycles from \(300\text{K}\) to \(673\text{K}\) with constant heating rate of \(3.33 \times 10^{-2} \text{ K s}^{-1}\).

Thermo-gravimetric analysis (TGA) were performed up to 693 K on samples submitted to the same heating used in MS to assess possible weight loss.

To assess possible structural changes occurring after heating-cooling cycles, X-ray diffraction (XRD) experiments were performed by using the Co K-\(\alpha\) radiation (\(\lambda = 1.78897 \text{ Å}\))
Experiments at high temperature

From about 300 to 373 K (black curve) E modulus exhibits an increasing trend while $Q^{-1}$ (grey curve) behaves in the opposite way.

Heating up to 373 K affects only residual water present in the pores. If the samples are cooled down from 373 K and rewetted, original characteristics are restored thus the process is reversible.

Above 373 K a broad and asymmetric $Q^{-1}$ maximum at 523 K is observed. E modulus decreases and exhibits two slope changes around 473 and 573 K. The asymmetric shape of $Q^{-1}$ maximum is due to the overlapping of two different maxima.

These phenomena are not present during cooling to room temperature and successive cycles.
TGA: weight loss with two stages corresponding to the ascending (1) and descending (2) parts of the $Q^{-1}$ maximum.
X-ray diffraction

Peaks of the mineral phases:
- hydroxylapatite $\text{Ca}_5(\text{PO}_4)_3(\text{OH})$
- fluorapatite $\text{Ca}_5(\text{PO}_4)_3\text{F}$
- chlorapatite $\text{Ca}_5(\text{PO}_4)_3\text{Cl}$
- carbonate-apatite $\text{Ca}_5(\text{PO}_4, \text{CO}_3)_3\text{F}$

XRD spectrum does not substantially change after the heating-cooling cycles
\[ Q^{-1} \propto \rho l^4 \omega \]

\[ \frac{\Delta G}{G} \approx -\beta \rho l^2 \]
Granato-Lücke string model

The model describes the changes of dynamic modulus and $Q^{-1}$ of metals in terms of string density and mean distance between pinning points.

$$Q^{-1} \propto \rho l^4 \omega$$

$$\frac{\Delta G}{G} \approx -\beta \rho l^2$$

$\beta$ constant,
$\rho$ density of vibrating strings,
$l$ average string length,
$\omega/2\pi$ vibration frequency
The concept that **energy loss** and **modulus** depend on the number of oscillating strings and on their mean length is of general validity and can be used to explain the anelastic phenomena of dentin.
Damping behaviour can be explained by considering the specific structure of dentin. **Fibres, fibrils** and **molecular chains** of the collagen the helix structure **are like strings** able to oscillate under an external applied stress.

The strings are **subjected to a complex system of constraints**, i.e. fiber-fiber bonds, fibril-fibril bonds and helix structure, which are **progressively modified as temperature increases** with a consequent effect on the anelastic properties of the organic phase.
• The ascending part of maximum can be ascribed to the loss of water located in the gaps between fibres and between fibrils.
• Despiralization of helix structure of collagen molecule occurs so molecules become free to oscillate and to contribute to the damping.
• The **increased number of oscillating strings** leads to $Q^{-1}$ increase and $E$ decrease.
• When temperature further increases (above 523 K) the water loss regards that inside the fibrils, i.e. water that guarantees the continuity of the peptide chain, so fibrils degrade and the chain is broken in fragments. Damping decreases owing to the shorter **length of the vibrating strings**.
• At 573 K combustion occurs further contributing to decreasing the damping and producing the second slope change of modulus.

\[
\frac{\Delta G}{G} \equiv -\beta \rho l^2
\]

\[
Q^{-1} \propto \rho l^4 \omega
\]
Conclusions

APPLICATION OF:
• FIMEC INSTRUMENTED INDENTATION
• MECHANICAL SPECTROSCOPY

ALLOW US TO DETERMINE:

1. PROPERTIES AND BEHAVIOUR OF HUMAN DENTINE such as

   - The presence of a broad and asymmetric $Q^{-1}$ maximum at 523 K.
   - Two evolution stages, which basically correspond to the ascending (stage 1) and descending (stage 2) parts of the maximum.

1. To relate those results to loss of fluids and degradation of collagen.
Thank you for attention