Peter R. Hoskins · Patricia V. Lawford
Barry J. Doyle *Editors*

Cardiovascular Biomechanics

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Preface

This book is concerned with cardiovascular biomechanics; this is the study of the function and the structure of the cardiovascular system using the methods of mechanics. It has become clear that this area lies at the heart of all the major cardiovascular diseases such as atherosclerosis and aneurysms; diseases which are responsible for some one-third of world's deaths. The underpinning principle which will be referred to several times in this book is that the cardiovascular system adapts in order to normalise its own mechanical environment. The cardiovascular system is able to do this because mechanical forces are sensed by tissues, and deviations from 'normal' result in biological changes which affect structure. The study of cardiovascular biomechanics therefore requires an interdisciplinary approach involving biology, medicine, physics, engineering and mathematics. This book is an introductory text suitable for students and practitioners in all these different fields. The book is suitable as a textbook to accompany a final-year undergraduate or masters (M.Sc.) course with roughly one or two lectures per chapter. It is also suitable as a first text for researchers and practitioners in cardiovascular biomechanics. The book is divided into four main sections; introductory Chaps. 1–2, Chaps. 3–8 on biomechanics of different components of the cardiovascular system, Chaps. 9–13 on methods used to investigate cardiovascular biomechanics (in clinical practice and research), and Chaps. 14–17 written from a perspective of diseases and interventions. There are two appendixes; one with questions for each chapter (multiple-choice questions, short-answer and long-answer questions), one with a glossary of 900+ terms. In order that the book is accessible by a mixed audience the text concentrates on explanations of physical principles without the use of complex mathematics. A few simple equations are used and there are no derivations of equations. The book is heavily illustrated with examples drawn from modern investigative techniques including medical imaging and computational modelling.

Cardiovascular biomechanics is a field that continues to evolve. Each chapter includes a number of key references so that the interested reader can use this book as a bridge to the research literature.

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Contents

Chapter 1
Introduction to Solid and Fluid Mechanics

Peter R. Hoskins

Learning outcomes

1. Explain the difference between a solid and a fluid.
2. Describe features of stress–strain behaviour of a solid measured using a tensile testing system.
3. Explain stress–strain behaviour of biological and non-biological materials in terms of their composition.
4. Define Young's modulus.
5. Describe the measurement of Young's modulus using a tensile testing system.
6. Discuss values of Young's modulus for non-biological and biological materials.
7. Define Poisson ratio and discuss values for different materials.
8. Describe viscoelasticity, its effect on stress–strain behaviour, and models of viscoelasticity.
9. Discuss linear elastic theory and its applicability to biological tissues.
10. Define hydrostatic pressure and values in the human.
11. Define viscosity in terms of shear stress and shear rate.
12. Describe different viscous behaviours.
13. Describe measurement of viscosity.
14. Describe typical measures of viscosity for different fluids.
15. Discuss Poiseuille flow: pressure-flow relationships for flow of Newtonian fluid through a cylinder.
16. Discuss Reynolds number and flow states.
17. Discuss pressure-flow relationships in unsteady flow in cylindrical tubes.
18. Discuss energy considerations in flow including the Bernoulli equation.

An understanding of the functioning of the cardiovascular system draws heavily on principles of fluid flow and of the elastic behaviour of tissues. Indeed, much of the

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cardiovascular system consists of a fluid (blood), flowing in elastic tubes (arteries and veins). This chapter will introduce basic principles of fluid flow and of solid mechanics. This area has developed over many centuries and Appendix 1 provides details of key scientists and their contribution.

The concept of a fluid and a solid is familiar from everyday experience. However, from a physics point of view, the question arises as to what distinguishes a fluid from a solid? For a cubic volume element there are two types of forces which the volume element experiences (Fig. 1.1); a force perpendicular to a face and a force in the plane of a face. The forces perpendicular to the face cause compression of the material and this is the case whether the material is liquid or solid. The force parallel to the face is called a shear force. In a solid, the shear force is transmitted through the solid and the solid is deformed or sheared. The shear force is resisted by internal stresses within the solid and, provided the force is not too great, the solid reaches an equilibrium position. At the nano level the atoms and molecules in the solid retain contact with their neighbours. In the case of a fluid, a shear force results in continuous movement of the material. At the nano level the atoms and molecules in the fluid are not permanently connected to their neighbours and they are free to move. The key distinction between a fluid and a solid is that a solid can sustain a shear force whereas a fluid at rest does not.

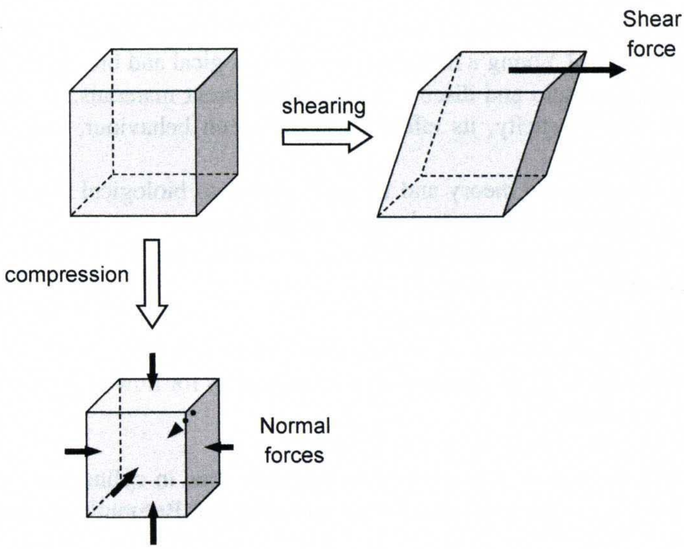

Fig. 1.1 A cube of material is subject to force parallel to a face which cause shearing and forces normal to each face which cause compression

1.1 Solid Mechanics

Solid mechanics is concerned with the relationship between the forces applied to a solid and the deformation of the solid. These relationships go by the name of the 'constitutive equations' and are important in areas such as patient-specific modelling discussed in Chap. 11. In general, these relationships are complex. For small deformations many materials deform linearly with applied force, which is fortunate as both experimental measurement and theory are relatively straightforward. This section on solid mechanics will start with 1D deformation of a material, develop linear elastic theory, then describe more complex features including those of biological materials.

1.1.1 1D Deformation

The elastic behaviour of a material is commonly investigated using a tensile testing system. A sample of the material is clamped into the system and then stretched apart. Both applied force and deformation are measured and can be plotted. Figure 1.2 shows the force-extension behaviour for steel. For many materials, such as steel and glass, the initial behaviour is linear; a doubling of applied force results in a doubling of the extension. In this region the material is elastic in that it will follow the same line on the force-extension graph during loading or unloading. The material is elastic up the point *Y*, which is called the 'yield point' but, after the yield point, the slope of the line decreases. The material is softer in that small changes in force result in large changes in extension. Beyond the yield point the material becomes plastic in that the material does not return to its original shape after removal of the force but is permanently deformed. In Fig. 1.2 further increase in force eventually leads to fracturing of the material at the point U, called the 'ultimate tensile strength' (UTS).

The force-deformation behaviour can be understood at the atomic level. The chemical bonds between atoms and molecules are deformable and small

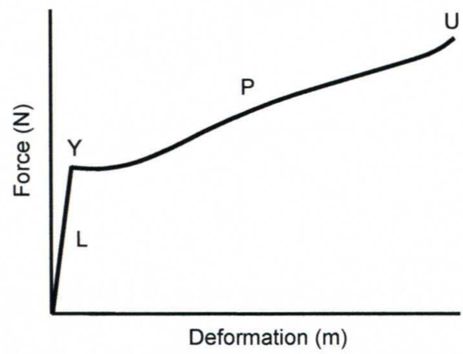

Fig. 1.2 Force-extension curve for steel. *L* linear behaviour; *Y* yield point; *P* plastic deformation; *U* (uniaxial) ultimate strength. Redrawn from Wikipedia under a GNU free documentation licence; the author of the original image is Bbanerje. https://commons.wikimedia.org/wiki/File:Hyperelastic.svg

deformations from the equilibrium position can be tolerated without change in structure. The equations governing the force-extension behaviour at the atomic level demonstrate linear behaviour and the macroscopic behaviour of a material is the composite of a multitude of interactions at the atomic and molecular level. In the plastic region there are changes in structure at the atomic and molecular level. In many materials this arises through slip processes involving the movement of dislocations or through the creation and propagation of cracks.

Biological materials are generally composite in nature. From a mechanical point of view the most important components are collagen fibres, elastin, reticulin and an amorphous, hydrophilic, material called 'ground substance' which contains as much as 90 % water. The elastic behaviour of the biological tissue is determined by the proportion of each component and by their physical arrangement. For example, collagen fibres in the wall of arteries are arranged in a helical pattern. Collagen is especially important in determining mechanical properties of soft biological tissues. Collagen is laid down in an un-stretched state. These unstressed fibres have a wavy, buckled shape, referred to as 'crimp'. On application of a force, the fibres begin to straighten and the 'crimp' disappears and, as a result, the tissue deforms relatively easily. With increasing extension the fibres straighten fully and resist the stretch. This leads to collagen having a non-linear force-extension behaviour, which explains the non-linear force-extension behaviour of most biological soft tissues.

A simple 1D tensile testing system can also be used to demonstrate viscoelasticity. It was stated above that in elastic behaviour the loading and unloading curves are the same. For a viscoelastic material they are different. In elastic behaviour the application of a force results more or less immediately in deformation of the material. Viscoelastic behaviour is associated with a time- lag between the applied force and the resulting deformation. The term 'viscoelastic' implies that the material has a mix of elastic and viscous properties. If the tensile testing system stretches the material in a cyclic manner, then as the tissue is loaded and unloaded, the resulting force-deformation curve will be in the shape of an ellipse (Fig. 1.3). During loading the force increases but the extension increases more slowly. During unloading the

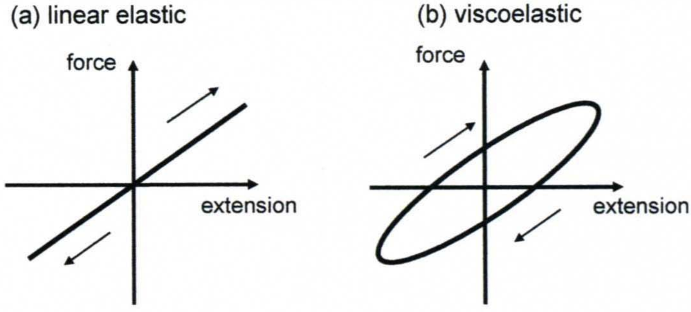

Fig. 1.3 Force-extension curves for cyclically varying force. **a** For a pure linear elastic material the loading and unloading curves are identical. **b** For a viscoelastic material the loading and unloading curves are different and are part of a loop

force decreases but the extension decreases more slowly. If the viscous component is low compared to the elastic component then the loading and unloading curves will be close together. For materials with a higher viscous component the curves are more separated and the width of the ellipse is larger.

1.1.2 Young's Modulus

In Sect. 1.1.1 the discussion of elastic behaviour was in terms of applied force and deformation. However the quantities stress and strain are more widely used in theory and experiment. The stress, σ, is the force, F, per unit area, A, and has units of pascals. The strain, ε, is the ratio of the extension, δl, divided by the original length, l, and is a dimensionless quantity.

$$\sigma = \frac{F}{A} \tag{1.1}$$

$$\varepsilon = \frac{\delta l}{l} \tag{1.2}$$

The Young's modulus, E, is a measure of the elastic behaviour of a material and is a fundamental mechanical property. Young's modulus is the ratio of stress divided by strain (Eq. 1.3). The units of E are pascals (Pa) or newtons per square metre (N m^{-2}).

$$E = \frac{\sigma}{\varepsilon} \tag{1.3}$$

Young's modulus is commonly measured using a tensile testing system. The value E is equal to the slope of the line on the stress–strain plot. For a linear elastic material the slope is constant over much of the range of stress/strain and the mechanical properties of the material may be described by a single value of E. For non-linear materials such as rubber or soft biological tissues, the value of E is dependent on the strain. For such materials the 'incremental elastic modulus' may be defined as the change in stress over the change in strain over a small section of the stress–strain curve (Eq. 1.4).

$$E_{\text{inc}} = \frac{\Delta\sigma}{\Delta\varepsilon} \tag{1.4}$$

Figure 1.4 shows the Young's modulus of a number of common materials. Note that the scale is logarithmic with a range of 9 orders of magnitude. Hard materials such as ceramics, metals and glasses have very high values of elastic modulus. These are usually quoted in gigapascals (GPa). Wood and wood products have lower values of elastic modulus, but still have a very wide range from very hard

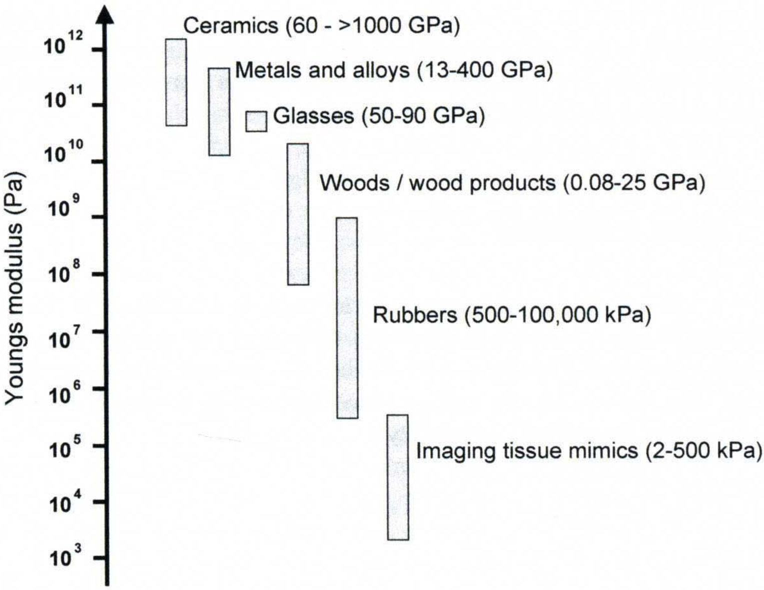

Fig. 1.4 Young's modulus *E* of common materials

woods such as oak, to very soft woods such as balsawood. Rubbers also have a very wide range from the hard vulcanised rubber used in tyres to the soft silicone rubber used in baby's dummies. The lowest elastic moduli values on the graph are for materials that mimic soft tissue used in phantoms for testing medical imaging systems. These are designed to mimic key properties of soft biological tissues, such as fat and muscle, and have low elastic modulus values in the range 2–500 kPa. Figure 1.5 shows the Young's modulus of a number of different biological tissues, taken from Sarvazyan et al. (1998). Again, there is a huge range of values. Bone and tooth enamel have the highest values of elastic modulus; liver, muscle and fat the lowest values.

The observant reader might have noted that it has been stated that the constitutive equations for soft biological tissues are complex and that the stress–strain behaviour is non-linear. How then is it justified in reporting Young's modulus, which generally applies to simple materials with linear stress–strain behaviours? This question will be addressed in Sect. 1.1.8; after more complex constitutive models have been considered.

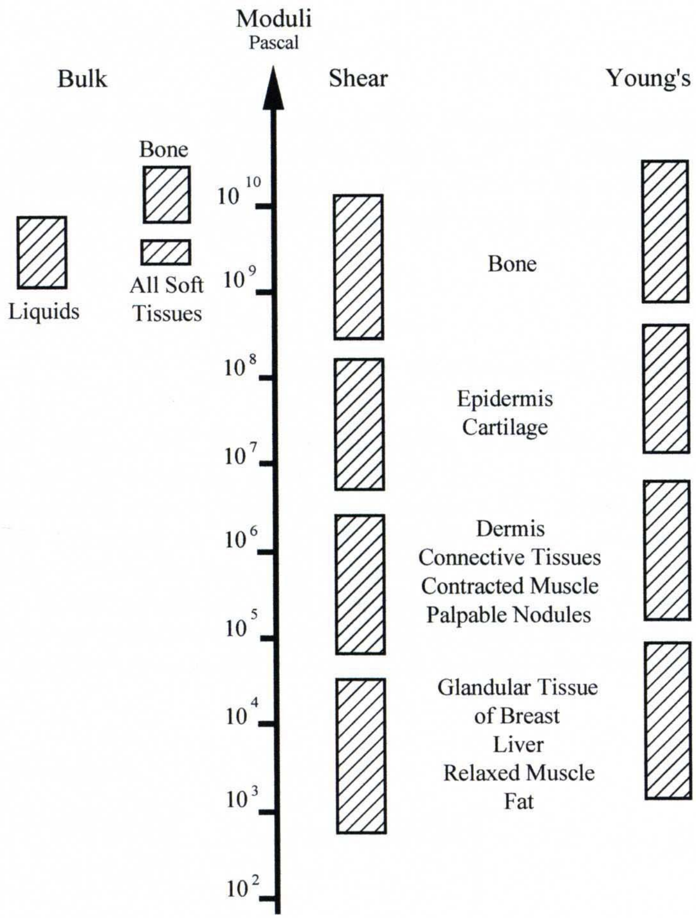

Fig. 1.5 A summary of data from the literature concerning the variation of the shear modulus, Young's modulus and bulk modulus for various materials and body tissues. Reproduced from Sarvazyan et al. (1995), with permission of Springer

1.1.3 Poisson's Ratio

When a material is stretched in the z direction there is usually compression in the x and y directions, and when a material is compressed there is usually expansion in the other two directions. This is called the Poisson effect. The Poisson ratio v is given by the fractional change in length in the x direction divided by the fractional change in length in the z direction (Eq. 1.5).

$$\upsilon = \frac{\delta x/x}{\delta z/z} \tag{1.5}$$

For incompressible materials, that is, materials where the volume does not change when loaded, the Poisson ratio has a value of 0.5. Soft biological tissues contain large amounts of water and have Poisson values close to 0.5. For many materials such as metals, glasses and concrete, the Poisson value is in the range 0.2–0.4.

1.1.4 Models of Viscoelastic Behaviour

Elasticity and viscosity can be represented by a spring and a dashpot. The spring responds immediately to being stretched, which represents the purely elastic behaviour of a material. For a dashpot, there is a delay between the stretching force and the extension, which represents the viscous behaviour of a material. A viscoelastic material can be represented as a combination of a spring and a dashpot and there are various configurations, three of which are shown in Fig. 1.6. These are; the Maxwell model where the spring and the dashpot are in series, the Voigt model where the spring and the dashpot are in parallel and a model consisting

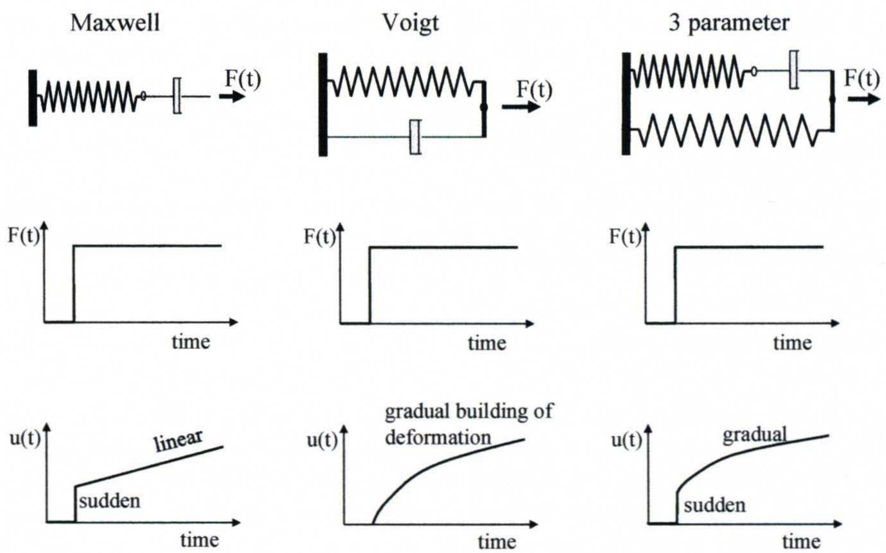

Fig. 1.6 *Top row* models of viscoelastic behaviour using combinations of a spring (elasticity) and dashpot (viscosity); Maxwell model, Voigt model, 3-parameter model. *Middle row* a sudden force is applied to the tissues. *Bottom row* The distension *u* is shown as a function of time for each model