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BASIC BIOMECHANICS *of the* MUSCULOSKELETAL SYSTEM

Margareta Nordin, P.T., Dr. Sci.

Director. Occupational and Industrial Orthopaedic Center (OIOC) Hospital fOf Joint Diseases Orthopaedic Institute Mt. Sinai NYU Health Program of Ergonomics and Biomechanics, New York University Research Professor Department of Orthopaedi" and Environmental Heallh Science School of Medicine. New York University New York. New York

Victor H. Frankel, M.D., Ph.D., KNO

President Emeritus Hospital for Joint Diseases OrthopaedIC Institute Professor of Orthopaedic Surgery New York University School of Medicine New York, New York

Dawn Leger, Ph.D., Developmental Editor

Kajsa Forssen, Illustrator

Angela Lis, P.T., M.A.. Editorial Assistant

$~~ LipPINCOIT WILLIAMS & WILKINS

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foreword

dents in orthopaedics Ihal during the past 10 years have llsed the texl. This book is written for stu- dents and with a major input from students and will hopefully be used to edunHe students and res- idents for many ~Iears to come. Although the basic information contained in the book remains largely unchanged, a considerable amount of extra infor- mation has been provided throughoul. ~Ve have also made a special point to document with the key references any significant changes in the field of biomechanics and rehabilitation.

Mechanics and biology have always fascinated humankind, The irnportance or understanding the biomechanics of the musculoskeletal system cannot be underestimated, Much a([entian has been paid in recent years to genetic and biomo- leClilar research, but the stlld~' of the mechanics of structure and of the whole body s}'stcm is still of immense importance. Musculoskeletal ail- ments arc among the most prevalent disorders in the wodd and will continue to grow as the pop- ulation ages.

It has always been m)' interest lO bridge the gap between engineering knowledge and clinical carc and praclice, This book is written primarily for clinicians such as orthopaedists, physical and occupational therapists, clinical ergono- misls, chiropractors, and olher health profes- sionals who arc acquil-ing a working knowledge of biomechanical principles for use in the evalu- ation and treatment of musculoskeletal dysfunc- tion. We only hope that if you find this book in- t~resting. you will seek more in-depth study in the field of biomechanics. Enjo\' it. discuss it, and become a beller clinician and/or researchcl: Vve are extremely proud that *Basic Biome- clUluics oj" the !\tlllscliioskeic/lli Sysle111* has been designated "A Classic" by the publishers, Lippincott Williams & Wilkins. We Ihank the readers, students, professors, and all who ac- quire thc tcxt and lise it.

*VielO,.* H. *Frallkel, M.D., Ph.D., KNO*vii

Since the days when I first studied biome- chanics in Sweden with Carl Hirsch, through my years as an orthopaedic surgeon, teacher, and re- searcher, I have alway's emphasized combining basic and applied research with clinical experi- encc, This text represents my fifth effort to inte- grate biomechanical knowledge into clinical training for patient carc. It is not a simple task but by relating the basic concepts of biomechan- ics to everyday life, rehabilitation. orthopaedics. traumatology, and patient care are greatly en- hanced. Biomechanics is a multidisciplinary spe- cialty, and so we have made a special effort to in- vite contributors from many disciplines so that individuals from dilTerent fields may feel com- Fortable reading this book.

Together with an invaluable team, Margareta Nordin and I have produced Ihis third edition of *Basic Biol1lechanics oFthe A'lusclt!o.\'keletal Systelll,* The new edition is shall1cncc! and improved thanks to the input from the students and resi·

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Biomechanics uses physics and engineering con- cepts to describe the motion undergone by the various body segments and forces acting on these body parts during normal activities. The inter-relationship of force and motion is impor- tant and must be understood if rational treat- ment programs are to be applied to muscu- loskeletal disorders. Deleterious effects may be produced if the forces acting on the areas with disorders rise 10 high levels during exercise or other activities of daily living.

The purpose of this text is to acquaint the read- ers with the force-motion relationship \vithin the musculoskeletal system and the various tech~ niqucs llsed to understand these relationships. The third edition of *Basic Biol1/eclwllics of rite iHllScliloskeleral System* is intended for use as a textbook either in conjunction with an introduc- tory biomechanics course or for independent study. The third edition has been changed in many ways, but it is still a book that is designed for use by students who are interested in and want to learn about biomechanics. It is primarily written for students who do not have an engi- neering background but who want to understand the most basic concepts in biomechanics and physics and how these apply to the human body

Input from students has greatly improved this third edition. We have used the book for 10 years in the Program of Ergonomics and Biomechanics at New York University', and it is the students and residents who have suggested the changes and who have continuously shown an interest in de- veloping and irnproving this book. This edition has been further strengthened by the contribu- tion or the students over the past year. \Vc formed focus groups to understand better what thc stu- dents wanted and applied their suggestions wher-

**Preface**

ever possible. We retained the selected examples to illustrate lhe concepts needed for basic knowl- edge of the musculoskeletal biomechanics; we also have kept the important engineering con- cepts throughout the volume. We have added four chapters on applied biomechanics topics. Patient case studies ancl calculation bo:'\cs have been added to each chapter. *\Ne* incorporated flowcharts throughout the book as teaching tools. The text will serve as guide to a deeper under- standing of musculoskeletal biomcchanics gained through funher reading and independent research. The information presented should also guide the rt.'ader in assessing the literature on biomechanics. "Vc have attemptcd to provide therapcutic exam- ples but it was not our purpose to cover this area; instead, \ve have described the undel'lying basis for rational therapcutic or exercise programs.

An introductory chapter describes the inlpor- lance of the study of biomechanics, and an ap- pendix on the international system of measure- ments serves as an introduction to the physical measurements used throughout the book. The reader needs no more (han basic knowledge of mathematics to fully comprehend the material in the book, but it is important to review the ap- pendix on the Sl System and its application to biomechanics.

The body of the third edition is then divided into three sections. The first section is the Bio- mechanics of Tissues and Stnlcturcs of the Mus- culoskeletal System and covers the basic biome- chanics of bone, ligaments, cartilage. tendons, muscles, and nenres. The second section covers the Biomechanics-of Joints, including every joint system in the human body. Chapters range from the foot and ankle through the cervical spine, and co\'er eveI:" joint in between. The third sec-

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tion cover~ some topics in Applied Biomechan- ics, including chapters on fracture fixation; arthroplasty; silting, standing and lying; and gait. These arc basic chapters that sl:l\re to intra· c1uce topics in applied biomechanics: they arc not in-depth explorations of the subject.

Finally. we hope that the revision and expan- sion of this third edition of" *Basic 13io11leclulJIics*

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*oFthe kluscu/oskelela/ Syslel1l* will bring about an increased awareness of the imparlance of bio- mechanics. II has never been our intention to complL'tely cover the subject, but instead provide a basic introduction to the field that will lead to further study or this important lopic.

*Margarela NOf(!;11 alld Viclor H. Frankel*

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This book was made possible through the out-

JUlian boxes. Angela look this book to her heart, standing contributions of many individuals. The

and \ve arc all the bettcr for her passion and chapter authors' knowledge and understanding of

attention to detail. the basic concepts of biomechanics and their

The illustrator: Kajsa Forssen. has now worked wealth of experience have brought both breadth

on all three editions of this text. Her never-failing and depth to this work. Over the past 10 years.

grasp of hiomechanical illustrations, her simpli. questions raised by students and residents have

city and exactness of figures, is always appre- made this book a better teaching tool. The Third

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Our colleagues al the Occupational and A book of this size with its large number of

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Diseases Orthopaedic Institule functioned as editor, Dawn Leger's continuous effort and

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*klargareta Nordin and Victor fl. Frankel*

**Contributors**

Gunnar B. J. Andersson, M.D., Ph.D.

Craig J, Della Valle, M.D. Professor and Chairman

NYU-HJD Department of Orthopaedic Surgery Department of Orthopaedic Surgery

Hospital for Joint Diseases Rush-Presbyterian-SI. Luke's Medical Center

School of Medicine Chicago, IL

New York University

Thomas P. Andriacchi, Ph.D.

New York, NY

Biomechanical Engineering Division

Victor H. Frankel, M.D., Ph.D., KNO Stanford University

President Emeritus Stanford, CA

Hospital for Joint Diseases Orthopaedic Institute

Sherry I. Backus, M.D., P.T. Senior Research Physical Therapist and Research Associate Motion Analysis laboratory

Professor of Orthopaedic Surgery New York University School of Medicine NevI York, NY

Hospital for Special Surgery

Ross Todd Hockenbury, M.D. New York, NY

River City Ortl1opaedic Surgeons

Ann E. Barr, Ph.D., P.T.

LouiSVille, KY

Assistant Professor

Clark T. Hung, Ph.D. Physical Therapy Department

Assistant Professor College of Allied Health Professionals

Department of Mechanical Engineering and Center for Temple University

Biomedical Engineering Philadelpllla, PA

Columbia University

Fadi Joseph Bejjani. M.D.. Ph.D.

New York, NY

Director of Occupational Musculoskeletal Diseases

Debra E. Hurwitz. Ph.D. Department

Assistant Professor University Rehabilitation Association

Department of Orthopaedics Newark, NJ

Rush·Presbyterian-St. Luke's lvIedical Center

Maureen Gallagher Birdzell, Ph.D.

Chicago, IL

Departmenl of Orthopaedic Surgery

Laith M. Jazrawi. M.D. Hospital for Joint DiseasesiMI. Sinai NYU Health

NYU·HJD Department of Orthopaedic Surgery New York, NY

Hospital for Joint Diseases

Marco Campello, P.T., M.A. Associate Clinical Director Occupational and Industrial Orthopaedic Center

School of Medicine New York University New York, NY

Hospital for Joint DiseasesiMI. Sinai NYU Health

Frederick J, Kummer, Ph.D. New York. NY

Associate Director, Musculoskeletal Research Center

Dennis R. Carter. Ph.D. Professor Biomechanical Engineering Program Stanford University Stanford, CA

Hospital for Joint DiseasesiMt. Sinai NYU Health Research Professor, NYU-HJD Department of Orthopaedic Surgery Scl100l of Medicine New York University New York, NY ~.- -\_. \_\_.\_---.\_.\_-\_.\_--\_.\_ \_.\_.\_.- \_.\_-\_.\_.\_.\_-\_. \_.-..\_.-.\_-\_ \_.\_.\_-\_.\_ \_ \_-\_.\_-- \_ \_\_ \_.\_.

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Dawn Leger, Ph.D. **Adjunct Assistant Professor** NYU-HJD Department of Orthopaedics School of Medicine **New York University** New York, NY

Jane Bear-Lehman, Ph.D., OTR, FAOTA **Assistant Professor of Clinical Occupational Therapy Department of Occupational Therapy Columbia University College of Physicians and Surgeons** New York. NY

Margareta Lindh, M.D., Ph.D. **Associate Professor Department of Physical MeeJicine and Rehabilitation** Sahlgren Hospital **Gothenburg University** Gothenburg, Sweden

Angela Lis, M.A., P.T. Research Physical Therapist **Occupational and Industrial Orthopaedic Center** Hospital for Joint DiseasesiMt. Sinai NYU Health New York, NY **Associate Professor** Physical Therapy Program **(orporacion Universitaria Iberoamericana** Bogota, COLOMBIA

Tobias Lorenz, M.D. **Fellow** Occupational and Industrial Orthopaedic Center **Hospital for Joint Diseases/Me Sinai NYU Health** New York, NY

Goran Lundborg, M.D. **Professor** Department of Hand Surgery **Lunds University** Malmo Allmanna Sjukhus Malmo, Sweden

Ronald Moskovich, M.D. **Associate Chief** Spine Surgery NYU-HJD Department oj Orthopaedic Surgery **Hospital for Joint Diseases** School of Medicine **New York University** New York, NY

Van C. Mow, Ph.D. **Director** Orthopaedic Research Laboratory Department of Orthopaedic Surgery **Columbia University** New York, NY

Robert R. Myers, Ph.D. **Associate Professor Department of Anesthesiology University of California San Diego** La Jolla, CA

Margareta Nordin, P.T., Dr. Sci. **Director, Occupational and Industrial Orthopaedic Center** (OIOC) **Hospital for Joint Diseases Orthopaedic Institute** !vlt. Sinal NYU Health **Program of Ergonomics and Biomechanics New York University Research Professor** Department of Orthopaedics and Environmental Health **Science School of Medicine, New York University New York, NY**

Kjell Olmarker, M.D., Ph.D. **Associate Professor Department of Orthopaedics** Sahlgren Hospital **Gothenburg University Gothenburg, Sweden**

Nihat bzkaya (deceased) **Associate Professor Occupational and Industrial Orthopaedic Center Hospital for Joint Diseases Research Associate Professor Department of Environmental Medicine New York University NelN York, NY**

Lars Peterson, M.D., Ph.D. Gruvgat 6 **Vastra Frolunda** Sweden

Mark I. Pitman, M.D. **Clinical Associate Professor** NYU-HJD Department of Orthopaedic Surgery School of lvIedicine **New York University** New York, NY

Andrew S. Rokito, M.D. **Associate Chief. Spons Medicine Service ASSistant Professor** NYU-HJD Department 'Of Ortllopaedic Surgery School of Medicine **New York University** New York, NY

Bjorn Rydevik, M.D., Ph.D. **Professor and Chairman** Department of Orthopaedics Sahlgren Hospital **Gothenburg University** Gothenburg. Sweden

G. James Sammarco, M.D. **Program Director** fellowship in Adult Reconstructive Surgery foot and Ankle Orthopaedic Surgery Program The Center for Orthopaedic Care, Inc. Volunteer Professor of Orthopaedic Surgery Department of Orthopaedics **University of Cincinnati Medical Center Cincinnati, OH**

Chris J. Snijders, Ph.D. Professor Biomedical Physics and Technology faculty of Medicine **Erasmus University** Rotterdam, The Netherlands

CONTRI~uioRS,;:,' ~V.

*Steven* Stuchin, M,D. **Director Clinical Orthopaedic Services Director Arthritis SefYice Associale Professor** NYU-HJD Department of Orthopaedics **School of Medicine New York University** New York, NY

Shira Schecter Weiner, M.A., P.T. Research Physical Therapist **Occupational and Industrial Orthopaedic Center Hospital for Joint Diseases/Mt. Sinai NYU Health** New York. NY

Joseph D. Zuckerman, M.D. **Professor and Chairman** NYU-HJD Department of Orthopaedic Surgery **Hospital for Joint Diseases** School of Medicine **New York University** New York, NY

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BASIC BIOMECHANICS of *the* MUSCULOSKELETAL SYSTEM

**Introduction to Biomechanics: Basic Terminology and Concepts**

*Nihat OZkaya, Dawn Leger*

Introduction

Basic Concepts

Scalars, VeCtors, and Tensors Force Vector Torque and Moment Vectors Newton '$ l.aws Free-Body Diagrams Conditions for Equilibrium Statics Modes of Deformation Normal and Shear Stresses Normal and Shear Strains Shl?ar·5train Diagrams Elastic and Plastic Deformations Viscoelasticity Material Properties Based 011 Stress-Strain Diagrams Principal Stresses Fatigue and Endurance

Basic Biomechanics of the Musculoskeletal System

Part I: Biomechanics of Tissues and Structures Part 11: Biomechanics of Joints Part III: Applied Biomechanics

Summary

Suggested Reading

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I

*Introduction*

Biomechanics is considered a branch of bioengi- neering and biomedical engineering. Bioengineer- ing is an interdisciplinary field in which the princi- ples and methods from engineering. basic sciences. and technology arc applied to design. test, and man- ufacture equipment for use in medicine and to un- derstand, define, and solve problems in physiology and biology!. Bioengineering is one of several spe- cialty areas that corne under the general field of bio- medical engineering.

Biomechanics considers the applications of clas- sical mechanics 10 the analysis of biological and physiological svstems. Different aspects of biome- chanics utilize different parts or applied mechanics. For example, the principles of statics havc been ap- plied to analyze the magnitude and nature of forces involved in various joints and muscles of the nUls- culoskeletal system. The principles of dynamics have been utilized for motion description, gait analysis, and segmental motion analysis and have many applications in sports mechanics. Thc mc~ chanics of solids provides the necessary tools for developing the field constitutive equations For bio- logical systems that are used to evaluate their func- tional behavior under dilTerent load conditions. The principles of fluid mechanics have been used to in- vestigate blood flow in the circulatory system, air flow in the lung, and joint lubl'ication.

Research in biomechanics is aimed at improving our knowledge of a vcry complex structure-the hu- man body. Research activities in biomechanics can be divided into three areas: experimcntal studies, model analyscs, and applied research. Experimental studies in biomechanics arc done to determine the mechanical properties of biological materials, in~ eluding the bone, cartilage, muscle, tendon, liga- ment. skin, and blood as a whole or as parts constituting them. Theoretical studies involving mathematical model analy1ses have also been an im~ ponant component of research in biomechanics. In general, a model that is based on experimental find- ings can be used to predict the efrect of environ- mental and operational factors without resorting to laboratory experiments.

Applied research in biomechanics is the applica- tion of scientific knowledge to bcnefit human bc- ings. vVe know that musculoskeletal injury and ill- ness is one of the primary occupational hazards in industrialized countries. By learning how the mus- culoskeletal system adjusts to common work concli- tions and by developing guidelines to assure that

manual work conforms more closely to rhe physical limitations of the human body and to natural body rnO\'cmCnlS, these injuries rnay be combatlcd.

*Basic Concepts*

Biomechanics of the musculoskeletal system re- quires a good understanding of basic mechanics. The basic terminology and concepts from mechan- ics and physics arc utilized to clcscribe intcrnal forces of the human body. The objective of studying thcs~ forces is to understand the loading condition of soft tissues and their mechanical responses. The purpose of this section is to rC\'jew the basic con- cepts of applied mechanics that are used in biome- chanical literature and throughout this book.

SCALARS, VECTORS, AND TENSORS

Most of the concepts in mechanics arc either scalar or vector. A scalar quanlity has a magnitude only. Concepts such as mass, energy', power, mechanical work, and temperalure are scalar quantities. For ex~ ample, it is suffkicnt to say that an object has 80 kilograms (kg) of mass. A vector quanlity, con- versely, has both a magnitude and a direction asso- ciated \vith it. Force, moment, velOcity, and acceler- ation arc exall'lples of vector quanlities. To describe a force fully. one must state how much force is ap- plied and in which direction it is applied. The mag- nitude of a vector is also a scalar quantity. The mag- nitude of any quanlity (scalar or vector) is always a positi\'c number corresponding to the numerical measure of that quantity\_

Graphically, a vector is represented by an arrow. The orientation of the alTow indicates the line of ac~ tion and the arrowhead denotes the direction and sensc of the vectm: 'If 1110re than one vector must be shown in a single drawing, the length of each arrow must be proportional to the Inagnitude of the vector it represents. Both scalars and vectors arc special forms of a more general category of all quantities in mechanics called tensors. Scalars arc also known as "zero-01"(Ic1· tensors," whereas vectors aJ'e "first· order tensors." Concepts such as stress and strain, conversely, are "second-order tensors."

FORCE VECTOR

Force can be defined as· mechanical disturbance 01- load. Whcn an object is pushed or pulled, a force is applied on it. A force is also applied when a ball is3

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thrown or kicked. A force <:\Cling on an object may deform the objecl, change its slale of m~lion, 0"1' both. Forces may be c1assif-lcd in variolls ways ac- cording to their effects on the objects to \vhicf'l they arc applied or according to their orientation as con~­ pared with one another. For example, a force may be internal or external, normal (perpendicular) 0'1' tangential; tensile. compressive. 01" shear; gravita- tional (weight); or frictional. Any two or more forces acting on (\ single body 111ay be coplanar (acting on a lwo~dimcnsional plane surface); collinear (have a common line of action); concurrent (lines of action intersecting at a single point); or parallel. Note that weight is a special form of Force. The weight of an object on Earth is the gravitational force exerted bv the Earlh wcighl on the of mass an object of that on Ea~·t11 object. is Thc equal magnitude t; the mass ~'r

of the object times the magnitude of the gravita- tional acceleration, \vhich is approximately 9.8 mt> ters pCI' second squared *(111/s1).* For object weighs approximately 98 newtons exampl~,a (N) 10-k<J o~ Earth. The direction of weight is always vertically do\vl1\vard.

TORQUE AND MOMENT VECTORS

The effect of a roree on the object it is applied upon depends on how the rorce is applied and how the object is suppo!"ted. For example, when pulled. an open door will swing about the edge along which it is hinged is the torque lO the generated wall. \-Vh'll by eallses the applied the door force 10 swinabou~ cJ

an axis that passes through the hinges of the door. If one stands on the free-end of a diving board, the hoard will bend, What bends the board is the mo- mel1l of the body weight about the fixed end of the board. In general. torque is associated with the ro~ tnlional and twisting action of applied forces, while moment is related to the bending action. However, the mathematical defmition of moment and torque is the same.

Torque and moment arc vector quantities. The magnitude of the tonlue Of rnoment of a force about a point is equal to the mannitude of the force times the length of the shortc:t distance be- which tween the is known point and as the the lever line or of action moment of arm. the force

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Definition of torque. *Reprinred with permission from DZkaya, N. (998). Biomechanics. In w.N. Rom,* Environmental and Occupationa( Medicine *(3rd ed., pp,* 1437~1454), *New York: Lippincott·Raven,*

sider a person on an exercise apparatLls who is holding a handle that is attached to a cable (Fig. I-I), The cable is wrapped around a pulley and at- tached to a weight pan. The weight in the \veight pan stretches the cable such that the magnitude F of the tensile force in the cable is equal 10 thc weight of the weight pan. This force is transmitted to the person's hand through the handle, At this in- stant, if the cable allached to the handle makes an angle *0* with the horizontal. then the force E ex- erted by the cable on the person's hand also makes an angle *0* with the horizontal. Let 0 be a point on the axis of rolation of the elbow joint. To dcter- mine the magnitude of the moment due to force f about 0, extend the line of action of force f and drop a line from 0 that cuts the line of action of F at right angles. If the point of intersection or the twO lines is Q, then the dbtance d between 0 and Q is the lever arm, and thc magnitude of the rno~ ment M of force E about the clbow joint is M = dE The direction of the moment ,·cctor is perpendicu- lar to thc plane defined by the line of action of E and line 00, or for thb two-dimensional case, it is counterclockwise.

NEWTON'S LAWS

Relatively few basic laws govern the relationship betwcen applied forces and corresponding mo- tions, Among these, the laws of mechanics intro- duced by Sir Isaac Newton (1642-1727) are the most important. NeWLOn's first law states that an object at rest will remain at rest 01' an object in mo- tion will move in a straight line with constant ve- locity if the net force acting on the object is zero. Newton's second law states that an object with a nonzero net force acting on it will accelerate in the direction of the net force and that the magnitude of the acceleration will be proportional to the magni- tude of the net force. Newton's sccond law can be formulated as E = m ;), Here, E is the applied force. m is the mass of the object, and i! is the linear (translational) accelcration of the object on which the force is applied. If more than one force is acting on the object. then E represcnts the net or the re- sultant force (the vector sum of all forces), Another way of stating Newton's second law of motion is M = I Q., where M is the net or resultant moment of all forces acting on the objcct, I is thc mass moment of inertia of the object, and ~ is the angular (rota- , tional) acceleration of the object. The mass m and mass moment of incrtia I in these equations of mo- tion arc measures of resistance to changes in 1110-

lion. Th~ larger the inertia of an object, the more difficult it is to sel in motion or to SlOp if it is ai- rend)' in motion.

Newlon's third law states that to every action there is a reaction and that the forces of action and reaction between interacting objects are equal in magnitude, oppositc in direction, and have the same line of action. This law has important applica- tions in constructing free·body diagrams.

FREE·BODY DIAGRAMS

Free-body diagrams are constructed to help identify the forces and moments acting on individual parts of a system and to ensure the correct use of the equations of mechanics La analyze the system. For this purpose. the parts constituting a system are iso- lated from their surroundings and the effects of sur- roundings arc replaced by proper forces and mo- ments.

The human musculoskeletal system consists of many parts that are connected to one another through a cornplcx tendon. Iigamcnt, muscle, and joint SU·uctUfC. In somc analyses, the objective may be to investigate the forces involved at and around val'ious joints of the human body for different pos- tural and load conditions. Such analyses can be car- ried out by separating the body into two parts althe joint of interest and drawing the free-body diagram of one of the parts. For example, consider the arn1 illustrated ill Figure I~2. Assume thal the forces in- volved at the elbow joint arc to be analyzed. As il- lustrated in Figure 1-2, lhe entire body is separated into two at the elbow joint and the free-body dia- gram of the forearm is drawn (Fig. *1-2B).* Here,

E is the force applied to the hand by the handle of the cable attached to the weight in the weight pan,

\V is the total wcight of thc lower arm acting at the center of gravity of the lower arm,

£,\\1 is the force excrted by' the biceps on the ra- dius,£.,,; is the force exerted bv the brachioradialis muscles on the radius.

£.\l~ is the force exerted by the brachialis musclcs on the ulna, and f1 is the resultarll reaction force at the humero- ulnar and humeroradial joints of the elbow. Note that the muscle and joint reaction forces represent the mechanical effects of the upper ann on the lower arm. Also note that as illustrated in Figure 1-2;\ (which is not a complete free-body diagram). equal magnitudc but opposite muscle and joint re- action forces act on the upper arm as wcll.

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Forces involved at and around the elbow joint and the free-body diagram of the lower arm. *Reprinted with permission from dzkaya. N. (7998). Biomechanics. In W.N. Rom,* Environmen- tal and Occupational Medicine *(31d ed., pp. 1437-1454). New York: Lippincott-Raven.*

CONDITIONS FOR EQUILIBRIUM

Statics is an area within applied mechanics that is concerned with the anal~:sis of forces on rigid bod- ies in equilibrium. A rigid body is one that is as- sumed to undergo no deformations. [n reality, evcr)' object or matcrial may undergo deformation to an extent when acted on by forces. (n some cases, the amount of deformation may be so smalllhal il may not affect the desired analysis and the object is as- sumed to be rigid. In mechanics, the term cquilib-

dum implies that the body of concern is either at rest or moving with constant velocity. For a body to be in a slate of equilibrium, it has to be both in translational and rotational cquilibl-iul11. A body is in translational cquilibriun1 if the net force (vector sum of all forces) acting on it is zero. If the Ilt:t force is zero, then the linear acceleration (time rate of change of linear velocity) of the body is zero, or the linear velocity of the bod,Y is either constant or zero. A body is in rotational equilibrium if the net mo- ment (vector sum of the moments of all forces) act· ing on it is zero. If the net moment is zero, then the angular acceleration (time rate of change of angular velocity) of the body is zero, or the angulal· yelocily of the body is either constant or zero. Therefore, for a body in a state of equilibrium, the equations of motion (Newton's second law) take the following special forms:

~E = 0 and ~rvl = 0

rt is important to remember that force and mo- ment arc vector quantities. For example, with re- spect to a rectangular (Cartesian) coordinate sys- tem, force and moment vectors may have components in the .'\. y, and z directions. Therefore, if the net force acting on an object is zero, then the sum of forces acting in each direction must be equal lo zero (IF, = 0, IF, = 0, IF, = 0). Similarly, if the net moment on an object is zero. then the sum of moments in each direction must also be equal to zero (lM, = 0, lM,. = 0, lM, = 0). Thel'efore, for three-dimension force systems there arc six cOl1cii- lions ofequilibrium. For two-dimensional force sys- [ems in lhe xy-plane, onl~: three of these conditions (IF, = 0, ~F, = 0, and ~M, = 0) need to be checked.

STATICS

The principles of slatics (equations of equilibrium) can be applied to investigate the muscle and joint forces involved at and around the joints for various postural positions of the human body and its seg- ments. The immediate purpose of static analysis is to provide answers to questions such as: What ten- sion must the neck extensor muscles exert on the head to support the head in a specined position? \OVhen a person bends, what would be the force ex~ ertcd by the erector spinae on the fifth lumbar ver- tebra? Ho\\'" does the con1pression at the elbow, knee, and ankle joints vmy with externally applied forces and with different segmental arrangements? How docs the force on the femoral head vary with loads carried in the hand? \,Vhat arc the forces in~

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volved in various muscle groups and joints during different exercise conditions?

In general. the unknowns in static problems in- volving the musculoskeletal s:'stcm arc thc magni- tudes of joint reaction forces and muscle tensions. The mechanical analysis of a skelclal joint requires that we know the vector characteristics of tensions in the muscles, the proper locations of muscle at- tachments, the weights of body segmcnts, and the locations of the centers of gravity of the body seg- mems. Mechanical models are obviously simple representations of c0l11plex systems. Many models are limited by the assumptions that must be made to reduce the system under considcration to a st.ati- cally determinate one. Anv model can be improved by ~onsidering the comril)utions of other muscles, but (hat will increase the number of unknowns and make the model a statically indeterminate one. To analyze the improved model. the researcher would need additional information related to the muscle forces. This inforrrlalion can be gathered through electromyography measurements of muscle signals or by applying certain optirnization techniques. A similar analysis can be made to investigate forces involved at and around other major joints of the musculoskeletal system.

MODES OF DEFORMATION

When acted on by externally applied forces. objects may translate in the direction of the net force and rotate in the direction of the net torque acting on them. If an object is subjected to externally applied forces but is in stalic equilibrium. then it is most likely that there is some local shape change within the objec!. Local shape change under the effect of applied forces is known as deformation. The extent of deformation an object may undergo depends on many' factors, including the material properties. size, and shape of the object; environmental factors such as heat and humidity; and the nlagnitudc, di- rection, and duration of applied forces.

One way of distinguishing forces is by observing their tendencv to deform the object they are applied upon. For example. the object is said to be in ten- sion if the body tends to elongate and in compres- sion if it tends to shrink in the direction of the ap- plied forces. Shear loading differs from tension and compression in that it is caused b:! forces acting in directions tangent to the area resisting the forces ~ causing shear, whereas both tension and compres- sion are caused by collinear forces applied perpen- dicular to the areas on which they act. It *is* common

to call tensile and compressive forces normal or ax- ial forces: shearing forces are tangential forces. Ob- jects also deform when they are subjecled to forces that cause bending and torsion, which are related to the moment and torque actions of applied forces.

A matel"ial nwv respond differently to different loading configurations. For a given material. there may be different physical properties that must be considered while analyzing the response of that ma- terial to tensile loading as compared with compres- sive or sheai' loading, The mechanical properties of mnterials are established through stress analysis by subjecting them to various experiments such as uni- axial tension and compression, torsion, and bend- ing tests.

NORMAL AND SHEAR STRESSES

Consider the whole bone in Figure \-3;,\ that is sub- jected to a pair of tensile forces of magnitude F. The bone is in static equilibriulll. To analyze the forces induced within the bone, the method of sections can be applied by hypothetically cutting the bone into two pieces through a plane perpendicular to the long axis or the bone. Because the bone as a whole is in equilibrium, the two pieces must individually be in equilibrium as well. This requires that at the cut sec- tion of each piece there is an internal force that is equal in magnitude but opposite in direction to the externally applied force (Fig. 1-38). The internal force is distributed over the entire cross-sectional area of the cut section. and E represents the resultant of the distributed force (Fig. 1-3C). The intensity of this distributed force (force per unit area) is known as stress. For the case shown in Figure 1-3. because the force resultant at the cut section is perpendicular to the plane of the cut. the cOITesponciing stress .is called a normal or axial stress. It is customar:y to usc the symbol *(T* (sigma) to refer to normal stresses. As- suming that the intensity of the distributed force at the Cllt section is uniform over the cross-sectional area Aof the bone, then *u::::: FlA.* Normal stresses that are caused by forces that tend to stretch (elongate) matcl"ials aJ"C marc specincally known as tensile stresses; those that tend to shrink them are known as compressive stresses. According to the Standard In- ternational (SO unit system (see Appendix), stresses are measured in newton per square meter (N/m~), which is also known as pascal (Pa).

There is another form of stress, shear stress, which is a measure of the intensity of internal forces acting tangent (parallel) to a plane of cut. For

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BcDefinition of normal stress. *Reprinted wirh permission from OZkaya. N. (1998j. Biome- chanics.* In W.N. Rom, Environmental and Occupatiol1<11 r..,ledicine *(3rd ed., pp.*

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example. consider the whole bone in Figure 1-4A. The bone is subject to a number of parallel forces that act in planes perpendicular to the long a,is of the bone. Assume that the bone is cut into two parts through a plane perpendicular to the long axis of the bone (Fig. 1-48). If the bone as a whole is in equilibrium, its individual parts must be in equilib- rillm as well. This requires that there must be an in- ternal force at the cut section that ,lets in a direction tangent to the cut surFace. If the magnitudes of the external forces arc known, then the magnitude F of the internal force can be calculated by considering the translational and rotational equilibrium of onc of the parts constituting the bone. The intensity of the internal force tangent to the Clit section is known as the shear stress. It is customary to usc the symbol *T* (tau) to refer to shear stresses (Fig. 1-4C).

Assuming that the intensity of the force tangent to the area cut section is Aof the bone, uniform then *T* = over FlA.

the cross-sectional

NORMAL AND SHEAR STRAINS

Strain is a measure of the degree of deformation. As in the case of stress, two types of strains can be dis- tinguished. A norm~l'l strain is deflnecl as the ratio of the change (increase or decrease) in length to the original (undeformed) length, and is commonly de- noted with the symbol € (epsilon). Consider the whole bone in Figure \-5. The total length of the bone is I. If the bone is subjected to a pair of tensile forces. the length of the -bone may increase to I' or by an amount *.1i\* = I' -I. The normal strain is the ratio of the amount of elongation to the original

F,

F.,

Shear strains are related to distortions caused by shear stresses and arc cornmonly denoted with the symbol y (gamma). Consider the rectangle (ABCD) shown in Figure 1-6 thm is acted on by a pair of tan- gential forces that deform the rectangle into a par- allelogram (AB'C '0). 'If the relative horizontal dis- placement of the top and the bOllom of the rectangle is d and the height of the rectangle is h, then the average shear strain is the ratio of d and h. which is equal to the tangent of angle y. The angle y is lIsllall~" vcry small. For small angles. the tangent of the angle is approximately equal to the angle it- self measured in radians. Therefore, the average shear strain is "y = cllh.

Shear strains are related to distortions caused by shear stresses and arc cornmonly denoted with the symbol y (gamma). Consider the rectangle (ABCD) shown in Figure 1-6 thm is acted on by a pair of tan- gential forces that deform the rectangle into a par- allelogram (AB'C '0). 'If the relative horizontal dis- placement of the top and the bOllom of the rectangle is d and the height of the rectangle is h, then the average shear strain is the ratio of d and h. which is equal to the tangent of angle y. The angle y is lIsllall~" vcry small. For small angles. the tangent of the angle is approximately equal to the angle it- self measured in radians. Therefore, the average shear strain is "y = cllh.

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Definition of shear stress. *Reprintecl with permis- sion {rom Ozkaya, N, (I 99B). Biomechanics.* /11 *W.N. Rom,* Environmenlal and Occupational Medicine *(3rd ed" pp.* 1437-/454). *New York: Lippincorr- Raven.*

length, or E = *c,11* 1. If the length of the bone in- creases in the direction in which the strain is cal- culated. then the strain is tensile and positive. If the length of the bone decreases in the direction in which the strain is calculated, then the strain is compressive and negative.

Different I11mcrials may demonstrate different stress-strain relationships. Consid(~r the stress- strain diagrarn shown in Figure 1-7. There arc six distinct points on the curve, which arc labeled as O. P, E, Y, U, and R. Point 0 is the origin of the Sli'ess- strain diagram, which corresponds to the initial (no load, no deformation) state. Point P represents the proportionality limit. Between 0 and P. stress and strain are linearly proportional and the stress- strain diagram is a straight line. Point E represents the clastic limit. Point Y is the .\"ield point, and the stress *(T..* corresponding to the yield point is called the yield slrength of the material. At this Slress level, considerable elongation (yielding) can occur without a corresponding increase of load. U is the highest stress point on the stress-strain diagram. The stress (rll is the ultimate strength of the mater- ial. The last point on the stress-strain diagram is R, \vhich represents the nq)ture or failure poinl. The stress at which lhe failure occurs is called the rup- ture strength of the material. For some materials, it may not be easy to distinguish the elastic limit and the yield point. The yield strength of sLieh materials is determined by the offset method, which is ap- plied b.y drawing a line parallel to the linear section of the stress-strain diagram that passes through a strain level or approximately *0.2 % •* The intersection of this line with the stress-strain *ClWVC* is taken to be the vielel point, and the stress corresponding to this po-int is called the ~\pparent yield strength of the material.

Different I11mcrials may demonstrate different stress-strain relationships. Consid(~r the stress- strain diagrarn shown in Figure 1-7. There arc six distinct points on the curve, which arc labeled as O. P, E, Y, U, and R. Point 0 is the origin of the Sli'ess- strain diagram, which corresponds to the initial (no load, no deformation) state. Point P represents the proportionality limit. Between 0 and P. stress and strain are linearly proportional and the stress- strain diagram is a straight line. Point E represents the clastic limit. Point Y is the .\"ield point, and the stress *(T..* corresponding to the yield point is called the yield slrength of the material. At this Slress level, considerable elongation (yielding) can occur without a corresponding increase of load. U is the highest stress point on the stress-strain diagram. The stress (rll is the ultimate strength of the mater- ial. The last point on the stress-strain diagram is R, \vhich represents the nq)ture or failure poinl. The stress at which lhe failure occurs is called the rup- ture strength of the material. For some materials, it may not be easy to distinguish the elastic limit and the yield point. The yield strength of sLieh materials is determined by the offset method, which is ap- plied b.y drawing a line parallel to the linear section of the stress-strain diagram that passes through a strain level or approximately *0.2 % •* The intersection of this line with the stress-strain *ClWVC* is taken to be the vielel point, and the stress corresponding to this po-int is called the ~\pparent yield strength of the material.

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Strains arc calculated by dividing two quantities measured in units of length. For most applications, the deformations and consequently the strains in- volved may be very small (c,g" 0,001), Strains can also be gi\'en in percemages (e.g.. O.l%).

STRESS-STRAIN DIAGRAMS

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Definition of normal strain. *Reprinted with permission from 6zkaya. N.* (/998). *Biome- chanics.* In W.N. Rom, Environmental (1nd Occupational !v1edione *(lrd ed., pp.*

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Note that a given material may behave dilTcr~ ently under different load and environmental con~ ditions. If the curve shown in F'igurc 1~7 repre- sents the stress"strain relationship for a material under tensile loading, there ma).o' be a similar but different curve representing the stress-strain rela- tionship for the same material under compressive or shear loading. Also. temperature is known 10411- leI' the relationship between stress and strain. For some materials, the stress-strain relationship may also depend on the rate at which the load is ap- plied on the material.

ELASTIC AND PLASTIC DEFORMATIONS Elasticit:-.· is defined as lhe ability or a material to resume its original (stress-free) size and shape on removal of applied loads. 1n other words, if a load

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Definition of shear strain. *Reprinted wirh permis- sion from OZkaya, N.* (1998). *Biomechanics. In W.N. Rom,* Environmental and Occupational Medicine *(3rd ed.. pp.* /437-1454). *New York: Lippincou-Raven.* **•**

S' ~C /---------7 C' *r1.t*/ / /

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Stress-strain diagrams. *Reprinted with permission from Ozkaya, N.* (1998). *Biomechanics. In W.N. Rom,* Environmental and Occupational Medicine *(3rd ed.. pp.* 1437-1454)., *New York: Lippincou- Raven.*

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is applied on a material such that the Stress gener- ated in the material is equal to or less than th~ elastic limit, the deformations that took place in the material will be cOlllpletcl.v recovered once the applied lands arc removed. An elastic material \vhose stress·strain diagram is a straight line is called a linearly clastic material. For such a matc- the stress is linearly proportional to strain. slope of the stress-strain diagram in the e1as- region is called the elastic or Young's rnodulus of the material. which is commonly denoted by E. ,Therefore, the relationship between stress and strain for linearly elastic materials is *a* := E€. This equation that relates normal stress and strain is called a material function. For a given material. different material functions may exist for different modes or derormation. For example, SOme materi- als may exhibit linearly elastic belHwior under shear loading. For such materials, the shear stress *T* is linearly proportional to the shear strain y, and the constant of proportionality is called the shear modulus, or the modulus of rigidity. If G repre- sents the modulus of rigidity, then ,. = Gy. Combi- nations of all possible material functions for a given material form the constitutive equations for that material.

Plnsticity implies permanent deformations. Ma- terials may undergo plastic deformations follo\ving elastic deformations when they are loaded beyond their elastic limits. Consider the stress-strain dia- gram of a material under tensile loading (Fig.I-7). Assume that the stresses in the specimen arc brought to a level greater than the yield strength of the material. On removal of lhe applied load. lhe material will recover the elastic deformation that had taken place by following an unloading path par- allel to the initial linearly elastic region. The point where this path cuts the strain axis is called the plastic strain. which signifies the extent of perrl1a~ nent (unrecoverable) shape change that has taken place in the material.

Viscoelasticity is the characteristic of a material that has both fluid and solid properties. Most ma- terials arc classified as eilher fluid or solid. A solid material will deform to a ccrLain extent when an exlernal force is applied. A continuously applied force on a Ouid body will cause a continuous de- formation (also known as flow). Viscosity is n fluid property thut is a quantitative measure of rcsis· tance to flow. Viscoelasticity is an example of how areas in applied mechanics can overlap, because it ulilizes the principles of both fluid and solid me- chanics.

G

E

linearly elastic material behavior. *Reprinted* wirh *permission from OZkclycl, N.* (1998)\_ *Biomecll<lflics. In W.N. Rom.* Environmental and Occupattonal MecH- cme *(3rd cd., pp J437-1Li54.J. Ne....,; York: Lippincott- Raven*

VISCOELASTICITY

\·Vhcn they are subjected to relatively low stress lev- els, many materials such as metals exhibit elastic material behavior. They undergo plastic deforma- tions at high stress levels. Elastic materials deform instantaneously when they are subjected to exter- nally applied loads and resume their original shapes almost instantly when the applied loads are re- mo\·cd. For an elastic material, stress is a function of strain only, and the strcss-strain relationship is unique (Fig. 1-8). Elastic materials clo not exhibit time-dependent behavior. A different gl'OUp of mate- rials, such as polymer plastics, metals at high tem- peratures, and almost all biological materials, ex- hibits gradual deformation and recovery when subjected to loading: and unloading. Such materials are called viscoelastic. The response of viscoelastic materials is dependent on how quickly (he load is applied or removed. The extent of deformation that viscoelastic materials undergo is dependent all the rate at which the deformation-causing loads are ap- plied. The stress-strain relationship for a viscoelastic material is not unique but is a f1.lI1ction of time or the rate at which the stresse.s and strains are developed in the material (Fig. 1·9). The word "viscoelastic" is made of two words, Viscosity is a fluid property and

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suain Ii)rale with an instantaneous strain that would remain at a constant level until the load is removed (Fig. I-lOB). At the instant when the load is removecl, the deformation will instantl)' and completely recover. To the same constant loading condition, a vis- coelastic material will respond with a strain in- creasing and ciCCI-casing graduall)r. If the material is I".·I•?rviscoelastic solid, the recovery will eventually be complete (Fig. 1-.1 *DC).* If the material is viscoelastic

; '"'r...•. fluid, complete recovery will never be achicved and there will be a residuc of defOl'mation lerr in the material (Fig. 1-IOD). As illustrated in Figure

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*l-11A,* a stress·relaxation experiment is conducted

Strain rate~dependentviscoelastic material be- havior. *Reprinted with permission /rom 6zkaya. N. (1998). Biomedlc1llic5. In WN. Rom,* Environmental and Occupational Medicine *(3rd ed., P.o. 1.137-/454). New York: Lippincorr·Raven.*

is a measure of resistance to now. Elasticity is" solid material property. Therefore, viscoelastic materials possess both nuid- and solid-like properties.

For an elastic material, the energy supplied to deform the material (strain energy) is stored in the material as potential energy. This energy is avail- able to return the material to its original (un- stressed) size and shape once the applied load is re- moved. The loading and unloading paths for an elastic material coincide. indicating no loss of en- ergy. Most elastic materials exhibit plastic behavior at high stress levels. For e1asto-plastic materials, some of the strain energy is dissipated as heat dur- some ing plastic or the defat-mations. strain energy For is stored viscoelastic in the materials, material as potential energy and some of it is dissipated as heat regardless of whether the stress levels are small or large. Because viscoelastic materials ex- hibit lime-dependent material behavior. the differ- ences between elastic and viscoelastic material re- sponses are most evident under time-dependent loading conditions.

Several experimental techniques have been de- signed to analyze the time-dependent aspects of material behaviOl: As illustrated in Figure 1-1004, a

Creep and recovery test. *Reprinred wirh permis-* creep and recovery test is conducted by applying a

*sion from Ozkay,l, N.* (1998). *Biomechanics. In W.N.* load on the matcl¥ial, maintaining the loael at a con-

*Rom,* Environmental and Occupational Medicine stant level for a while, suddenly removing the load,

*(3rd ed., pp.* 1437-745/1). *New York: Lippincorr-* and obsen;jng the material response. Under a creep

*Raven.* and recovery test. an elastic material will respond

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'by straining the Olalcriallo a level and maintaining the constant strain while observing the stress re- sponse of the material. Under a stress-relaxation lcst, an elastic mater-ial will respond with a stress developed insw.ndy and maintained at a consWnl level (Fig. I-II *B).* That is, an elastic malcrial will not exhibit a stress-relaxation behavior. *t\* viscoelas- lie material. conversely', will respond with an initial high stress level that will decrease over time. If the m;terial is a viscoelastic solid, the stress level will nevcr rcduce to zcro (Fig, I-lie), As illuSlrated in Fiourc I-II *D,* the stress will evct11uallv reduce to o . zero for a viscoelastic nuid.

MATERIAL PROPERTIES BASED ON STRESS-STRAIN DIAGRAMS

The stress-strain diagrams of two or Il"wrc materials can be compared to determine \vhich m<:ucrial is rei· atively stiffer, l1C:lrdcl~ tougher, more ductile, or more brittle. For example, the slope of the stress-strain di~ agram in the clastic region represents the clastic modulus that is a measure of the relative stiffness of materials. The higher the elastic modulus, the stiffer the material and the higher its resistance to defor- mation. Aductile material is one that exhibits a large plastic deformation prior to failure. A britlie mater- ial, such as glass, shows a sudden failure (rupture) without undergoing a considerable plastic deforma- tion. Toughness is a measure of the capacity of a ma- terial to sustain permanent defonllation. The tough- ness of a matedal is measured b~: considering the total area under its stress-strain diagram. The larger this area, the tougher the malerial. The ability of a material to store or absorb energy without perma- nent deformation is called lhe resilience of the ma- terial. The resilience of a material is measured by its modulus of resilience, which. is equal to the area un- der the stress-strain curve in the elastic region.

Although thcy arc not directl\' rclated to the stress-strain diagrams, other important concepls are used to describe material properties. For cxam~ pie, a material is called homogeneous if its proper- ties do not vary from location to location within the material. A material is called isotropic if its proper- lies are independent of direction. A material is called incompressible if it has a constant denSity.

PRINCIPAL STRESSES

There are infinitely many possibilities of con- structing elements around a given point wilhin a structure. Among these possibilities, there may be

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0"0 = E,

0"0 = E,

B to aU: C 10 Gl1- D 10

Stress-relaxation experiment. *Reprinted with per- mission from Ozkaya, N.* (1998). *Biomechanics. In WN. Rom.* Environmental and Occupational Medicine *(Jrd ed.• P.o.* 1437-1454). Ne~··1 *York: Lippincott-Raven.*

one: element for which the normal stresses are maximum and minimum. These maxin1lrm and minimum normal stresses arc called the principal stresses, and the planes whose normals are in the directions of the maximum and minimum stJ"esses are called the principal plancs, On a principal plane, (he normal stress is either maximum or minimum. and the sheal" stress is zero. It is known that fracture or material failure occurs along the planes of maximum stresses, and structures must be designed by taking into consideration the max- / imulll stresses involved. Failure by yielding (ex- cessive deformation) n.lay occur whenever the largest principal stress is equal to the yield strength of the material or failure by rupture may

,o· 1f~[[HAPTER}1 •

occur whenever the largest principal stress ,is

IIp equal to the ultimate strength of the material. For a given structure and loading condition. the prin- cipal stresses ma~" be within the limits of opera- tional safely. However, the structure must also be checked for critical shearing stress. called the maximum shear stress. The maximum shear SlI-es$ occurs on a material element for which the normal stresses are equal.

A aomax am·- o - tI :··· - ..... "7" ..... - - ] - - ':'.~ • Tension - - - - .......- - - - - - -

(j" : ..' •••••

- ~:.;. - - - ";"~ - - ~:~

• • • lime (j min ••••• •••••

FATIGUE AND ENDURANCE 1cycle Compression

Principal and maximum shear stresses are useful in predicting the response of materials (0 static load- ing configurations. Loads that Illay not cause the failure of a structure in a single application may cause fracture when applied repeatedly. Failure may occur aher a few or many cycles of loading and un- loading, depending on factors such as the amplitude of the applied load, mechanical properties of the material, sIze of the structlire, and operational con- ditions. Fracture resulting from repeated loading is called fatigue.

Several experimental techniques have been de- veloped to understand the fatIgue behavior of ma- terials. Consider the bar shown in Figure 1-12;:1. Assume that the bar is made of a material whose ultimate strength is *U'w* This bar is first stressed to a mean stress level *(1m* and then subjected to a stress fluctuating over time, sometimes tensile and other times compressive (Fig. 1-128). The amplitude *(T:,* of the stress is such that the bar is subjected to a maxImum tensile stress less than the ultimate strength of the material. This reo versible and periodic stress is applied until the bar fractures and the number of cycles N to frac- ture is recorded. This experiment is repeated all specimens applying stresses having or the varying same material amplitude. properties A typical by

result of a fatigue test is plotted in Figure 1-12C on a diagram showing stress amplitude versus

The fatiguc behavior or a material depends on numbct· of cycles to failure. For a given N. the cor-

several factors. The higher the temperature in responding stress value is called the fatigue

which the material is used, thc lower the fatigue strength of the material at that nun1ber of cycles.

strength. The fatigue behavior is sensitive to surface For a given stress level, N represents the fatigue

imperfections nnd the presence of discontinuities amplitude life of the material. versus number For some or matel"ials, cycles curve the stress levels

B

°0- - - - - - - - - - - - - -

L-\_,---'---,---,--N c10' 10' 10' Fatigue and endurance. *Reprinred with permission from Olkaya, N.* (1998). *Biomechanics. In W.N. Rom,* Environmeniat and Occupational Medicine *(3rd ed., pp.* 1437-1454). *New York: Lippincou-Raven.*

within the material that can cause stress concentra- tions. The fatigue failure starts \vith the creation of off. The stress *CT,* at which the fatigue curve levels

a small crack on the surface of the material. which off is called the endurance limit of the material.

can propagate under the effect of repeated loads, re- Below the endurance limit, the material has a

sulting in [he rupture of [he material. high probability of not failing in fatigue, regard-

Orlhopaedic devices lII)dergo repeated loading less of how many cycles of stress are imposed on

and unloading as a result of the activities of the pa- the material.

tients and the actions of their 111uscles. Over a pe-

riod a' fi;.;:ati~n of vears. device a weight-bearing can prosthetic dc\·icc or be subjected to a consiclerabk number of cycles of stress reversals as a result or noHnal daily activity. This cyclic loading and un-

~~*.•l;"N* can cause faLigue failure of the device.

*Biomechanics the Musculoskeletal System*

even a simple task c.'\ecuted b.v the musculoskeletal svstcm requires a broad. in-depth :~;' knowledge of various fields that ma~' include 1110- tor control, neurophysiology, physiology. physics. and pose biomechanics. ancl intention or For a example, task and PART II: BIOMECHANICS OF JOINTS

based on the pur- the sensOl'~' infor- mation gathered from the ph~'sical cndronmcnl and orielllatioll of tilL' body and joints, the central ;~; nervous system plans a strategy for a task execu- :;.: lion. According to the strategy adoptc:d. Illuscles /..'.,' will be recruited lO provide Lh<..' forces and 1110- mcnts required for the movement and I.Jalance of the s.)'slem. Consequently, the internal forces will be changed and soft tissues will experience differ- ent The load purpose conditions.

or this book is to present a \\'cll~ balanced synthesis of information gatllCred frorn various disciplines. pro\'iding a basic understanding of biomechanics of the musculoskeletal system. The material presented here is organized (0 cover three areas of musculoskeletal biomechanics.

PART I: BIOMECHANICS OF TISSUES AND STRUCTURES

The material presented throughout this textbook provides an introduction to basic biolllechanics of the musculoskeletal system. Part I includes chap- ters on the biomechanics of bone. articular carti- lage, tendons and ligaments, periphcral nerves, and skeletal muscle. These are augmcnted wilh case studies to illustrate the imponarll concepts for un- derstanding the biomechanics of biological tissues.

Part II of this textbook covers the major joiots of the human body, from the spine to the ankle. Each chapter contains information about the structure and functioning of the joint. along with case studies illuminating the clinical di~lgnosisand management of joint injlll)' and illness. The chnpters are written

b~" clinicians (Q provide an introducLor~: level of knowh:dge about each joint s~'stem.

PART III: APPLIED BIOMECHANICS

A new section in the third edition of this book in~ troduces important issues in applied biomechanics. These include the biomechanics of fracture fixation; arlhroplasty; sitting, standing. and lying; and gait. It is important for the beginning studenl to under- stand the application or biomechanical principles in clirfcrcnt clinical areas.

*Summarv*

1 Biomechanics is a young and dynamic fidd of study based on the recognition thaI conventional cllginccl'ing thcorks and methods can be useful for understanding and solving problems in physiology and medicine. Biomcclwnics considers the applica- tions of classical rncchanics to biological problems. The flcld of biomechanil:s flourishes from the coop- eration among life scientists, physicians, engineers. and basic scientists. Such cooperation requires a certain amount of common vocabulary: an engineer must learn some anatom~:and ph)'siology, nnclmed- ical personnel need to understand some basic con- cepts of physics and mathematics.

2 The information presented throughout this textbook is drawn from a large scholarship. The au- thors aim to introduce some of the basic concepts of biolllechanics related to biological tissues and joints. The book does nOl intend to provide a com- prehensive review of the literature, and readers are encouraged to consult the list of suggcsted reading below to supplcmcnt theil' knowledge. Some basic textbooks arc listed here, and studcnts should con- sult peer-revicwed journals for in-depth presenta- Lions of the latest research in specialty arcas,

SUGGESTED READING Black. J. (19SS). Onhop'lcdic Bionmleriuls in R,'sl,.'ardl and Prac-

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**The System International d'Unites (51)**

*Dennis R. Carter*

The 51 Metric System

**Base Units Supplementary Units Derived UnitS**

Specially Named Units

Standard Units Named for Scientists

Converting to 51 From Other Units of Measurement

I

*The''S! Metric System*

,,;The System Intemational d'Unites (SI), the metric system, has evolved into the most exacting system of 'measures devised. In this section. the 51 units of lllcasurcmcnl used in the science of mechanics arc described. ences have SI been units omitted used for in electrical the sake or and simplicity.

light sci-

~\_\_ • *fifo'*

<*""'i,ilY* ' BASE UNITS

--~

A" The 51 units can be considered in three groups: I, tbe base units; 2. the supplementary unils; and 3. *'j;",*the derived units (Fig. App-I). The base units are a

DERIVED UNITS

,',small group or standard measurements lhal have been arbitrarilv defined. The base unit for length is '-'rhe meler (Ill), "and the base Llnit for Illass is the kilo- gram (kg). The base units for time and temperature are the second (s) and the kelvin (Kl. respectively. Definitions 01" the base units have become.:: increas· .ingly sophisticated in response to the expanding needs Hnd capabilities of the scientific community (Table App-I). For example, the meter is now de-

QUANTITIES EXPRESSED IN TERMS OF UNITS FROM WHICH THEY WERE DERIVED

fined in terms of the \V~\'c1cngth of radiation emit- ted from the krvpton-S6 atom.

SUPPLEMENTARY UNITS

ivlostunits of the 51 system are derived unils, mean- ing lhat they- are established from the base units in accordance with fundamcntnl physical principles. Some of these units are cxp['css~d in terms of the base units from which they are derived. Examples aloe area, pressed in speed, the Sf units and or acceleration. square meters which (m~). The radian (rad) is a supplemental)' unit (() measure plane angles. This unit, like the base units, is arbi- trarily defined (Table App-I). Although the radian is the 51 unit for plane angle. the unit of the degree has been retained for general use because il is firmly established and is widely Ltscd "round the world. A degree is equinllcnl to rrll80 rad.

arc ex- meters per (In/ssecond 2), respectively.

(m/s), and meters pCI' second squared

FORCE ACCELERATION OF MOMENT

FOACE

qp ,----.... newIon kg *m/s2*

speED

DENSITY

VOLUME

~DERIVED~ ~~ ;~:::::<':~:~~J"J ~~\ ~,I . \l/~ \' , ,~/~,:'!j //'/~'" *Il/'* \_~r;) AREA

,~m'?I~~'TS i~radian (fad)

------ PLANE ANGLE SUPPLEMENTARY UNIT

meter (m) kilogram (kg) LENGTH MASS

second (s) kelvin (K) TIME TEMPERATURE

BASE UNITS

The International System of Units.

19

DERIVED UNITS WITH SPECIAL NAMES

PRESSURE & STRESS pascal *N/m2*

ENERGY & WORK joule Nm

POWER watt *JIS*

TEMPERATURE degree Celsius K'- 273.15

I *Specially Nanzeel Units*

Other dedved units are similarl.v established from the base units but have been given special names (Fig through App-I the and lise Table of fundarnental App-I). These equations units are or defined physi- cal laws in conjunction with the arbitrarily defined Sf base units. For example, Newton's second law of motion states that when a body that is free to Il10vC is subjected to a force. it will experience ~m ~lCcelcr­ alion proportional to thai force and inversely pro- portional to its own mass. i\Jlathcmatically, this prin- ciple can be expressed as:

force = Illass X acceleration

The Sf unit of force, the newton (Nl, is lherefore defined in terms of the base SI units as:

1 N = 1 kg X I I11/S:! The Sf unit or pressure and stress is the pascal (Pa). Pressure is defined in hydroslaties as the force divided by the area of force application. Mathem4ll- icall~l, this can be expressed as:

pressure = force/area

~.\_.m..\_\_ ..\_ \_\_.\_\_.\_.\_\_.. \_

; IDefinitions *of* Sl Units

Base 51 Units

meter (01)

kilogram (kg)

second (s)

kelvin (k)

Supplementary SI Unit radian (rad)

Derived SI Units With Special Names

newton (N)

pascal (Pa)

joule (J)

wall (W)

degree Celsius (C)

The meIer is the length eqllal to 1,650,763.73 wavelengths in vacuum of the radiation corresponding to the transition belween the levels 2PHi and Sd" of the krypton-86 atom.

The kilogram is Ihe unit of mass and is equallO the mass of the international proto· type of the kilogram.

The second is the duration of 9.192,631,770 periods of Ihe radiation corresponding to the transition between the two hyperfine levels of the ground state of the cesium-133 atom.

The kelvin. a unit of thermodynamic temperature, is the fraction 1/273.16 of the ther- modynamic temperature of the triple point of water.

The radian is the plane angle between two radii of a circle that subtend on the circum~ ference of an arc equal in length to the radills.

The newton is that force \tvhich, when applied to celeration of 1 meter per second squared. 1N= a 1kg rpass rn/sof /.

1 kilogram, gives it an aC-

The pascal is the pressure produced by a force of 1 newton applied. with uniform dis- tribution, over an area of 1square meter. 1 Pa = 1 N/m~.

The joule is the work done when the point of application of a force of 1 newton is displaced through a dist~nce of 1 meter in the direction of the force. 1 J = 1 Nm.

The watt is the power that in 1 second gives rise to the energy of 1 joule. 1W ::: 1 J/s.

The degree Celsius is a unit of thermodynamic temperature and is equivalent to K - 273.15.

The 51 unit of pressure, the pascal, is therefore defined in terms of the base 51 units as:

I Pa = IN *I* I 111"

Allhough the S[ base llnil of temperature is the kc.:lvin, the derived unit of degree Celsius (OC 01' c) is much marc commonly used. The degree Celsius is equivalent to the kelvin in magnitude. but the ab~ solute value of the Celsius scale differs frol11 that of the Kelvin scale such that °C = K - 273.15.

,",Vhen the 51 s~'stcm is used in a wide variely of measurements, the quantities expressed in terms of the base. supplemental. or derived units ma~t be either very large or very small. For example, the arca on the head of a pin is an extremely small number when expressed in terms of square meters. Conversely, the weight of a whale is an extremely large number when expressed in terms of newtons. To accomrnodate the convenient representation of small or large quantities, a system of prefixes has been incorporated into the SI system (Table App-2), Each prefix has a fixed meaning and can be used with all 5Iunils. \!\Then used with the name or the unit, the prefb: indicates that the quanti!}! described is being expressed in some multiple or

Factors and Prefixes

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ten times the unit used. For example. the millime· 'tel' (mm) is used to represent one thousandth (10"·1) ---\_.\_-------- of one a meter billion A ampere

( coulomb O( degree celsius F farad H henry Hz hertz J joule

K kelvin

N newton fl ohm Pa pascal 5 siemens T testa

V volt

W walt

Wb weber

and a gigapascal (10') pascals.

(Gpa) is uscd to denotc

, SI Units Named After Scientists

Symbol Unit Quantity

electric current

electric charge

temperature

electric capacity

inductive resistance

frequency

energy

temperature

force

eleclrlc resistance

pressure/stress

electric conductance

magnetic flux density

electrical potential

power

magnetic flux

*Standard Units Nmned for Scientists* SI Prefix SI Symbol giga G

mega Iv1 kilo k

One of the more interesting aspects of the SI system is its lise of the names of famous scientists as st«n~ dard units. In each case, the lInit was named after a scientist in recognition of his contribution La the hecla h

Geld in which that unit plays a major role. Table deka da

App-3 lists a number of Slullits and the scientist for deci d

which each was named.

centi c rniJli rn

For example. the unit of force. the neWlon, was named in honor of the English scientist Sir Isaac Ncwlon (1624-1717). Hc wa' cducalcd 'II Trinil)' micro I' nana n

College College as at a Cambridge professor and or mathematics. later rclurned Early to Trinity in his pica p

career, Newton made fundamental contributions to malhcmalics Ihat formcd Ihc basis of diffcrcntial and integral calculus. His other major discoveries were in the fields of optics. astronomy. gravitation. and mechanics. His work in gravitation was pur- portedly spurred by being hit on the head by an ap- ple falling from a tree. It is perhaps poetic justice that the SI unit of one newton is approximately equivalent to the weight of a medium-sized apple. Newton W~lS knighted in t705 by Qucen i\ltary fot" his monumental contributions to science.

Scientist Country of Birth Oates Amphere, Andre·tvtarie France 1775-1836 Coulomb, Charles Augustin de France 1736-1806 (elsius. Anders Sweden 1701-1744 Faraday, Michael England 1791-1867 Henry, Joseph United States 1797-1867 Hertz, Heinrich Rudolph Germany 1857-1894 Joule. James Prescott England 1818-1889 Thomson, William (lord Kelvin) England 1824-1907 Newton, Sir Isaac England 1642-1727 Ohm. Georg Simon Germany 1787-1854 Pascal, Blaise France 1623-1662 Siemens, Karl Wilhelm (Sir William) Germany (England) 1823-1883 Testa. Nikola (roatia (US) 1856-1943 Volta, (ount Alessandro Italy 1745-1827 Watt, James Scotland 1736-1819, Weber, Wilhelm Eduard Germany 1804-189}.;;};'

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Conversion *of* Units

Length

1 centimeter (em) = 0.01 meter (01)

1 inch (in) = 0.0254 rn

1 foot (ft) = 0.3048 rn

1 yard (ydl = 0.9144 rn

1 mile = 1609 m

1 angstrom (A) = 10" rn

TimeI minute (min) :::; 60 second (s)

1 hour (h) = 3600 s

1 day (dl = 86400 s

Mass1 pound mass (1 brn) = 0.4536 kilogr~lm (kg)

1 slug ::: 14.59 kg

Force1 kilogram force (kgf) = 9.807 Nevvton (f\J)

1 pound iorce IIbO = 4448 N

I dyne (elyn) " I0 ~I

Pressure and Stress

Moment (Torque)

1 dyn-cm = i 0.7 N-rn

i Ibf-ft = 1.356 N-m

Work and Energy

1 kg-m' / s' = 1 N-m = 1 Joule IJ)

1 dyn-cm = 1 erg = lO-} J

1 Ibi-It = 1.356 J

Power

I kg· rn' / s' = 1 J/s = 1 Watt (1"1)

I horsepower (hpl = 550 Ibl·IVs = 7461"1

Plane Angle

J degree ("') .::;: 1</180 radian (fad)

1 revolution (rev) :=: 360"

1 *rev;:::* 211: raci -."-' 5.283 rad

Temperature

<>( '" "r~ ~ 273.2

1 *kgl* rn·s:' = 1 N/m" = 1 Pascal (Pal

I Ibi / in' (psi) = 6896 Pa

I Ibl / ft' (psO = 92966 Pa

I dyn / em' = O. I Pa

Reprinted with pefFnisslon from OZkaya. 1'1., & No~dJn. l'1i. (1999). Fundamentals oi Biomechanics: EqUllib· rium. Motion. and DeformatIon (2nd ed.) New York: Spnnger-verlag. p. 11.

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The unit of pressure and stress. the pascal. was named after the French physicist, mathematician, and philosopher Blaise Pascal (1623-1662). Pascal conducted important investigations on the <.:harnc- teristics of vacuums and barometers and also in- vented a machine that would make mathematical calculations. His work in the area of hydrostatics helped lay the foundation for the later developmenl of these scientific ficlds\_ In addition to his scicntific pursuits, Pascal was passionalely interested in reli- gion and philosophy and tllllS wrote extensively on a wide range of subjects.

The base unit of temperature, the kelvin. was named in honor of Lord Vv'illiam Thomson Kelvin (1824-1907). Named William Thomson, he was cd·

ucated at the Universit.\· of Glasgo\\" and at Cam- bridge University. Early in his career, Thol11son in- vcstigated the thermal propcnies of steam at a sci- entific laboratory in Paris. At thc age of 32, he returned to Glasgo\\" 10 accept the chair of Nalural Philosophy. His meeting with James JOllie in 1847 stimulated interesting discussions on the nature of heat, which eventually lecl to the establishment of Thomson's absolute scale of temperature, the Kelvin scale. In recognition of Thomson's contributions Lo the field of thermodynamics, King Edward VIl con- fen-cd on him the title of Lord Kelvin.

The commonly llsed unit of temperature, the de- gree Celsius, \\as named aftcr the Swedish as- lronomcl and lmenlO! Anders CelsiLl' (1701-1744).

'M'·'.'''T,r",

Celsius was appointed professor or astronomy at the Inii"p-",i,\' of Uppsala at the age of 29 and remained the university until his death 14 years latec [n 1742, he described the centigrade thermometcl- in a paper prepared for the Swedish Academ!' of Sci- The name of the centigrade temperature waS officially changed to Celsius in 1948.

*to Sf From Other of'Measurement*

Box App·t contains the formulae for the conversion ~f measurements expressed in English and non-Sl units into Sf units. One fundamental source of connJSlon in converting from one system to another is that (wo basic t~'Pes of Illeasun:menl systems exist. In the "ph\'sical" system (such as SI). the units of length. time, and *nUlSS* arc arbitrarily defined, and other units (including force) are derived fron1 these ,. base units. In "technical" or "gravitational" systems

(slich as the English system). the units of length, time, and *force* arc arbitrarily' defined, and other units (including mass) are derived from these base units. Because lhe units of force in gravitational svs- tems are in fact the *It'eights* of stan(~"\rd masses. c~n~ version to 51 is dependent on the acceleration of mass due to the Earth's gravity, By' international agreement. the acceleration due to gravity is *9.806650 m/s'.* This \'aille has been lIseel in establish- ing some of the conversion factors in Box App-I.

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I

Biomechanics of Tissue and Structures of the Musculoskeletal System

Biomechanics of Bone

*Victor H. Frankel, Margareta Nordin*

Introduction

Bone Composition and Structure

Biomechanical Properties of Bone

Biomechanical Behavior of Bone Bone Behavior Under Various loading Modes

Tension Compression Shear Bending Torsion Combined l.oading Influence of Muscle Activity on Stress Distribution in Bone Strain Rate Dependency in Bone Fatigue of Bone Under RepetitivE: Loading !nfluence of Bone Geometry on Biomechanical Behavior Bone Remodeling

Degenerative Changes in Bone Associated With Aging Summary

References

Flow Charts

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of *lJitijoduction*

the tissue. Bone scn·cs <.lS a rcsen/Oir for essential minerals in the bod~·. particularly calcium. TbJ:i~~rpose or the skeletal system is to protect in- te.r~.~V:"organs, provide rigid kinematic links and ~~ls:ctEtattachmcntsites, and facilitate Illuscle ac· :·:::;:~"pJ:tf91iYli~9,bodymovement.. Bone has unique stl·UC· '-:/:::"'{:<'Y~lit~I.-.rI1d mechanical properties that allow it to ,,:~,<,:;;::,/,):',si;(rrY()llt;these roles. Bone is among the body's ,::)},:;;ll~lrci~~tstrllctllres;onl)' dentin and enamel in the ::",';;::;.~/'i,1;i',.~t..?9Pp'.cally :,:;:::';:,:::':i:,..'./t~¢th.>qreharder. active It is one tissues of the in most the dynamic and body and re· >.";'\1.lAiri,S::~\ctiv0throughollt life. A highl~; vascular tis- s~e'.-.it tan ~lter has its an excellent properties capacity and configuration for self-repair in and rc- . spOI~~e" to changes in mechanical demand. For ex- ample, changes in bone dcnsit~1 arc commonly obsei~·ed after pL:riods of disuse and of greatly in- creased use; changes in bone shapL' are noted dur· iog fracture healing and after certain operations. .;: Thus, bone adapts Lo the mechanical demands

placed on il.

This chapter (iL'scribcs the composition and structure of bone tissue, the mechanical properties of bone, and the behavior or bone under different loading conditions. Various factors that affect the mechanical behavior of bone in vitro and in vivo

Bone minend is embedded in variousl~« oriented fibers of the protein collagen, the fibrous ponion of the extracellular matrix-the inorganic matrix. Collagen ribers (type l) are tough and pliable, yet they resist strelching ~lnd have lillie L'xtensibility. CollagL'1l composes ~\PPJ'().\.irnatdy *9(YYo* of the ex~ tracellular matrix and accounts for approximately 25 lO *3W>'r'* of the dr~! wl'ight of bone. A universal building block of the body, collagen also is the chie!" fibrous component of other skddal struc- tures. (A detailed descriplion 01" the microstnlcturc and mechanical behavior of collagen is provided in Chapters 3 and 4.)

The gelatinous ground substance slIlTounding thc mineralized collagcn fibers consists mainly of protein polysaccharides, or gl~"cosarninoglycans (GAGs), primarily in the form of cOlllplt:x macro· molecules called protcoglycans (PGs). The GAGs s~rvc as a cementing substance between layers of mineralized collagen fibers. Thcs~ GAGs, along wit.h various noncollagcI1ous glycoprotcins, consti- tute approximately 5% of the extracellular rnatrix. (The structure or PG:-:, whi,:h arc vital components of artie, i:il' '\_·;l~·{iL\gt·. is described in delail in Chap·

also are disclissed.

k r .) ..1

\Vater is fairl~: abundant in live bone. accounting *Bone Composition and Structure*

for up to *25%* of its total weight. Approximately *85°10* of the walt.'r is found in the organic matrix, around the collagen fibers and ground sllbstance, Bone tissue is a specialized connective tissue whose

and in the h.vdl·ation shells surrounding the bone solidcomposition suits it for its supportive and pro-

crysl~ds. The other 15% is localed in the canals and teclive roles. Like other connective tissues, it con·

cavities that house bone cells and carry nutrienls to sisrs of cells and an organic extraccllular matrix of

th~ bone tissuc. fibers and ground substance produced by the cells.

At the microscopic level, the fllndamcmal StI11C- The distinguishing reature of bone is its high con·

lllralunit of bone is the osteon, or haversian system tent of inorganic materials, ill the form of mineral

(Fig. 2·1). At the center of each osteon is a small salts, that combine intimately with the organic ma-

channel, called a haversian canal, lIlat conlains trix (Buckwalter et al., 1995). The inorganic compo·

blood vessels and nerVe fibers. The osteon itself con- nent of bone makes the tissue hard and rigid, while

sists of a concentric series of layers (lamellae) of the organic component giv~s bone its flexil)ility and

mineralized rnatrLx surrounding the central canal, a resilience. The composition of bone dirrers depend-

configuration similar to growth ring~ in a tree ing on site, animal age. dietar:.y histol)', and the pres·

trunk. ence or disease (Kaplan et aI., 1993).

Along the bOllndflries of each layel: or lamella, are In normal human bone, the mineral or inorganic

small cavities known as lacunae, each cOl1laining portion of bone consists primarily of calcium and

011(.' bone cell. or ostcocyte (Fig. 2-1 *C).* Numerous phosphate, mainly in the form of small crystals rc·

small channels, called canaliculi, radiate from each sembling synthetic hydroxyapatite crystals \vith

lacuna, connecling the lacunae or adjacent lamellae the composition Ca",(PO,)o(OI-l),. These minerals,

and ultimately reaching the haversian canal. Cell which a~count for 60 to 70% of ilS dry weight, give

processes extend from the osteocytes into the canali- bone its solid consistency. "Vatcr nccounts for 5 to

culi, allowing nutrients'[Tom the blood vessels in the *SOk* and the organic matrix makes tip the J'cmainder

haversian canal to reach the osteocyles.

.. , ..~l.'

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a~~l~¥!Tr:

Lamellae

OsteocYle~

Lacuna-----

B

A

A. The fine structure of bone is illustrated schematically in a section of the shaft of a long bone depicted with- out inner marrow. The osteom. or haversian systems. are apparent as the structural units of bone. The haver- sian canals are in the center of the osteons, which form the main branches of the circulatory network in bone. Each osteon is bounded by a cement line. One osteon is shown extending from the bone (20x). *Adapted from Basset!,* CAL. (1965). *£/ecrrical effects in bone.* SCI Am, 213.18. B, Each osteon consists of lamellae. concentric rings composed of a mineral matrix surrounding the

**•**

haversian canal. *Adapled irom Torrora G.J..* & *Anagnos- takas. N.P.* (198:1). Principles of Anatomy and Physiology *(4th edJ.* Ne~·v *York: Harper gRow.* C, Along the boundaries of the lamellae *are* small cavities known as lacunae, each of which contains a single bone cell, or osteocyte. Radi- ating from the lacunae are tiny canals, or canaliculi, into which the cytoplasmic processes of the osteocytes extend. *Adapted from Torrora G.;..* & *AOclgflosrak05. N.P.* (1984). Principles of AnalOmy and Physiology *(4th edJ. New York: Harper* & Ro~"/.

At the pcriphcl)! of each osteon is a cement line, a nrtrrow area of cement-like ground SubSlrtllCe composed primarily of GAGs. The canaliculi of Ihc osteon do not pass this cement line. Like the canali- culi, the collagen fibers in the bOl1e matrix intercon- nect from one larndla to another within an osteon but do not cross the cement line. This intertwining of collagen fibers within the osteon undoubtedly in-, creases the bone's resistance to mechanical stress and probably explains \_~vhy the cement line is the weakest portion of the bonc's microstructure.

A typical osteon is approximately 200 micronle- tcrs (J..l) in (lin·rnctcr. Hence, evclY point in the os- teon is no more than 100 J..llll from the centrally 10- catcd blood supply. In Ihe long bones, the osteons usually run longitudinally. but they branch [I'c- quelltly and anaslOmose extensively with each othcl:

Jnterstitial lamellac span the regions between complete O!"h:ons (Fig: 2-1..1). They arc continuous with the ostcons and consist of the same material in a difTerent geol11ctric configuralion. As in the os-

!1

teons, no point in the interstitiallamellae is farther than 100 JLm from its blood supply. The interfaces between these lamellae contain an alTa)' of lacunae in which oSleocytes lie and from which canaliculi cXlcnd.

!\l the macroscopic level, all Qonc,~ ..,!re..c.Qrnposed of twotvpes, ,00~os'se-qus,~isslle: cortical. or compact, bone ~\n~d ci\'nc-c-il~·~IS.-Ol:"tl'abeclllal~"'iJ()nc'(FE~-"-i~·2). COl'tiC,,1 hoi1"Tili'ms the Otlt",:,11-"II:or' cortex:of the b()I"1~,tll~Cl..has a ~l,~nse stt'llct,llre ,Si..11]i,ic.\I:t~)th~\i of ivory. C;ln~~Tlou's bo;;e within ihi~~I;ell is composed or-thin plates. o'j"lrabeclllae, in a loose mesll.-,~t1·.uc­ lUI~e; lhf([rlier'si-i·ccs--bctwe.en the lrabct:ulat:: are filled \\~hh red ;;all'O\';" (Fig. 2~j). C~I~~~ii~ll~-b~-~~~~e isaiTangeCi inconcentric lacunae-cQn'i"ainTilgJii!l)el- lae--ouiclocsn-ot em~l;-in haversian canals. The os- teoc:\;tes re'ceive nUlricn-islhrough canaliculi from blood vessels passing through the red marrow. ~.Q.r­ lieal..bone always SUITOt.II).~~~ G!.pccl!.ous lJ.one. bUl· thi-rdative quantily of each lype varies' among bones and wilhin individual bones according to funclional requirements.

Frontal longitudinal section through the head. neck, greater trochanter, and proximal shaft of an adult fe- mur. Cancellous bone, with its trabeculae oriented in a lattice, lies within the shell of cortical bone. *Reprinted with permission from Gray, H. (T* 985). Anatomy of the Human Body. *(73(h American ed.J. Philadelphia: Lea* & *Febiger.*

A, Reflected-light photomicrograph of cortical bone from a human tibia (40;<). B, Scanning electron pho- tomicrograph of cancellous bone from a human tibia *(30)-:). Reprinted wirh permission from Carrer. D.R.,* & *Nayes, We.* (1977). Compact bone fatigue damage. A microscopic ex- aminalion. Clin Onhop, 127, 265.

On a rnicroscopic level, bone consists of .woven and lamellar-bone (Fig. 2-4). Woven bOlle is-con-si(l- erecl"immatllrebone. This type or bone is rOllt1d i,,-the enlbryo.-in the newborn, in the- fracturc'callus, and in lhe riietaphysial region of growing bone as well as in tUIl-lOrS, ostcogcl]csis lmperfecla, and pagctic bone. Lal-nellar bone begins to form 1month arter birth and

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activelv replaces woven bon~. Lamellar bone is therC'- for~ a "marc" l11all~'c bOlle.---

All bones are §urrounded by a dC.llse \_\_ U\_bJ~olls membrane called the periosteulll (Fig. 2-1.-\). Its ouler fa)~ei: (~ -pe-i::nl-eat-e-(Ch~);-'-GToodvessels (Fig. 2-5) and nerve fibers that pass into the cortex via Volk,llunn's cH"nalS: c()r\-i1c-EUn·g~\\;i.t~-1 thc--rl~i\.;e,:sian eLlm\ls '\Il'(Cc',~,~~,6~,lir~~Lt(?tll~-,s,a.·!1·~,~,1..1('-li~,I)one..*f\* n in I'~<;-l~" o~\_~,~~?g,~n .i..~,J,\_~ver C()ll,t,,~1.'i Ils '" l?5)Il~ ~t:llsIY~ spo"nsible '(or ~(;nen~ting nc\v Gc;'n-c'''du'i-ing gro\Vtl~, .....", ,5:\_".". . .",.. "" •.'-""'-"""~""""""" -- """

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**-•**'--

-- -**•**--- " - Lamellar/'i ~ *.f*.\ *l*v~~, } *J* , ,~. \. \\)

Woven

Schematic drawing and photomicrographs of lamellar and woven bone, *Adapted from Kaplan, F.5., Hayes, We., Keaveny. T.M.,* er *a/.* (1994). *Form and funcrion of bone.* In *S.R. Simon (Ed.).* Orthopaedic Basic Science *(pp.* 129, *130). Rosemonr, fL: AAOS.*

and which long line";; th~cnli~ \VilTl'-y~cfl~~~~;~ ,'cRail'. t borles, heeen1ral m":Cco\;'cr'ccT'witTl -b0I1::'except (ostcoblasls). 'a- fall'; thin'llcr (;;;";;clujIatyj -iiuirio\\':~ "nl'e'moi~a-ne:llle-cndostcum, arlicufm- for The The' ~,,",;ilv:,\'Iiicll;~Jmed the- periosteum endostcum cartilage. joint-'Slld"a-ces, covers

InThe

co.n-.-- tai,'ls~~~s:l'~~;~l;ia~t; ,~,~d .alse),. giant n~-~,fli"'l-ll1-c'le:'l:t'~'a' l.ioli"-e"lisctlTfec! im pOI:f~li1i'I~(;fc's"'Ttl-dlc'l~cmodcling osteoela-sfs: both aMi1~ir"(:csoI'P of which play- ti 611 of bone.

'("'1::.;,

.~

Photomicrograph *FS.. Nayes, w,e..* Keaven}~ showing rA1.. the *et* vasculature M *(1994).* of *Form* cortical and *function* bone. *Adap,ed of bone. from* In *5.R,* K~lplan,

*Simon (Ed.).* Orthopaedic Basic Science *(p.* /3i}. *Rosemon£,* It: *AAOS.*

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*Biomechanical Properties ofBone*

Biomechanically, bone tissue may be regUl~ded a~ a t\v9~ph~,lse (biphasic) c01JlposiL(;~ I1),aterial;" with the mineral as onCjJhase and the collagen and ground substance as\_.the o~her. In such matcl"'ials (n nonbio~­ logic-al exal,'lple is fiberglass) in which a strong. brit- tle material is embedded in a w~akcc more Oexible one, the combined substances ;:tre stronger for their weight than is either substance alone (Bassett, 1965),

Functionally, the most important mechanical properties of bone are its strength and stiffness. These and other characteristics can best be under- stood for bone, or any other structure, by e:'\amin- ing its behavior under .loading, that is, under the in- nuence of externally applied forces. Loading causes a deformation, or a change in the dimensions, of the~-strllctllre.\·Vhen a load in a known direction is imposed on a structure, the deformation 01" that structure can be rncasurcd and ploued 011 a load- deformation curve. I\lluch information about the

strength, stiflness, and other mechanical properties of the structure can be gained b~' examining this curve.

A hypothetical load-deformation curve for a somewhat pliable fibrous structure, such as a long bone, is shown in Figure 2·6, The i1~~lial (straight line) portion of the curve, the c1asticJ'cgion. reveals the elasticity bf the structure, th:u is, its capacity for returning moved, As· io the its load original is applied, shape after deformati911 the load occurs is re-

btll is not permanent; th-e structure recovers its orig- inal sllape \\'hen unloaded. As loading c·ontinues. the outcrni"ost fibers of the struCture begin to yield at some point. This yidd poinl signals the elastic limit of the structure, As,the load exceeds this limit, the structure exhibits plastic behavior, I'cflecl~d in the second (curved) portion of the eurvc, the p!as,tic rc- gion. The structure \\~i11 no longer return to its origi- 11al dimensions when the load has been released; some residual deformation will be permanent. If loading-is progressively increased, the structure will fail at'somc point (bonc' will fracturc), This point is indic~-iled by the ultimate failure point on the CUI've,

Dro*S*

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Three parameters for determining the strength of a structure are reflected on the load-deformation Clll-ve: 1, the load that the structure can sustain be- fore failing; 2, the deformation ihat it cansustain before failing; and 3, the erler¥.Y that it CaIlstore be~ fore failing. The strel1gth in terms of loaclancLde- fm:"mation, or ultimate strength, is in,clicatedOllJhe curve by the ultimate failure point. The streI,lgtl1 in terms of energy storage is indicated by the size of lhe area under the entire cun'e. The larger the area, the greater the energy that build~upin tfle struc- ture as the load is applied. The stiffness of the structure is indicated by theslope of tl).<::.curve in the elastic region. Thesteepei::::the slope, the stifrer the material.

the load-deformation curve is useful for deter- Inining the mechanical properties of whole struc~ turessuch as a whole bone, an entire ligament or tendqn, or a metai implant. This knowledge is help- ful in the study of fracture behaviorand repail~ the response of a struetlJre to physical stress, or the er~ fect of various treatment programs. However, char~

cPlastic region i '" Yield

Ultimate point

failure point

Deformation / *1'-* D *1111* / Energy *1111111*AD'

Load-deformation curve for a structure composed of a somewhat pliable material. If a load is applied within the elastic range of the structure (A to B on the curve) and is then released, no permanent defor- mation occurs. If loading is continued past the yield point (B) and into the structure's plastic range (B to C on the curve) and the load is then released, perma- nent deformation results. The amount of permanent deformation that occurs if the structure is loaded to point 0 in the plastic region and then unloaded is represented by the distance between A and D. If loading continues within the plastic range, an ulti- mate failure point (C) is reached.

acterizing a bone or other structure in terms of the material of \vhich it is composed, independent of its geometry, requires standardization of the testing conditions and the size and shape of the test speci- mens. Such standardized testing is useful for com- paring the 111echanical properties of two or more materials, such as the relative strength of bone and tendon tissue or the relative stiffness of various ma- terials used in prosthetic implants. More precise units of measurement can be used when standard- ized samples are tested-that is, the load per unit of area of the sample (stress) and the amount of defor- mation in terms of the percentage of change in the sample's dimensions (strain). The curve generated is a stress-strain curve.

Stress)s:the load, or force, per unit area that de- velops on a plane surface within a struc:lll['e.)11 re- sponse'iO'exteI~ililll)'applied loads. The three ~111its most commonly used for measuring stress in stan- dardized samples of bone are ne\vtons per centime- ter squared (N/cm:;); newtons per meter squared, or pascals (N/m 2,Pa); and megancwtons per meter sqmii;cd; or megapascals (MN/m 2, MPa).

Strain is theclef()l'mation (change in dimension) that develoP?~yithiI1.~ls:tTuctureirl"I:t::sponse to ex- ternallyapplied loads. The two basic types of strain are ii'near strain, which causes "a- chang~ ill the lengtl;"'6f'dlcspecimen, and shear strain, which causes }i"'~I~~i:~¥?i~.(ll~angulari'·clationships \vithin the structure. Linea-I'· strain is measured as the amoliiltofflIlear de[ormati()n (lengthening or short- ening}"6fthesiilnple di\'ided by the sample's original length. It is a nondim<;nsj()llal paramel<;r expressed as a percentage (e.g:, centimeter per centimeter). She'~1'1:'stt'~lTI1"ismeasured as the all"1Q.~lIlL.ofangular ch~~!!.g,~..,,,(Y)\_i,l'-adglif\_\_~~lj:~leI)'il1gi"I"i.t}l<; pl.nne.oLin- terest in the sample. It is expressedil"il'actians (one radian-'e(ill~llsai)proximately57.3°) (International, Society of Biomechanics, 1987).

Stress and strain values can be obtained for bone by placing a standardized specimen of bone tissue in a testing jig and loading it to failure (Fig. 2~7). These values can then be plotted on a stress-strain curve (Fig. 2-8). The regions of this curve are simi- lar to those of the load-deformation cUll/e. Loads in the elastic region do not cause perll1aD.<;nL~I~J()rma­ lion, buC6ncc:the yield point is excee.deeJ,s()ll1c de- Formation L5 permanent. The strengthoftheJl1~lt.~r~ inl in terms of energy' storage is repre~ent~elby the area .~II1~1.<::I·theyntirecurve. The stifFness is repre- sented h.ytheslope of th<? curve in the~Iasticregion. A value for stiffness is obtained by dividing the

stress at ~\_\_point ir~\_.thc clastic (straight line) porti()11 of trlC"cllI"\'e by i·he-~t~,~~·in ~t that point. This value -is caITedtTle-nl0dlll~I~~ o!:-elastfcity (Young's modulus). Young'S modulus (E) is d~,T;'ed from Ihe relalionship betw';-en str"ss *('T)* an~.strain(~):

E=<r/E The elasticity 01" a material or the Young's modulus E is equal to the slope of the stress (<r) and strain (E) diagram in the clastic linear region. E represents the stiffness of Ih" material, such Ihat Ihe higher the elastic modulus or Young's modulus, the stiffer the material (Ozkaya & Nordin, 1999).

Mechanical propcrties differ in the two bone types. Cortical\_bq\_\_,!\_c ~~ s~.Hfer\_tlla!,! canccllQus\_ bone, withstanding greater stress but less strain before falllll·c. C:-1-n~c~ITol-i-s -bolic-rn·-\'itl:-O--~~·~I~-~;jn lip to *50%* of strains before yielding. \vhile cortical ~)(?nc viclds and fractures when the strain exccec!s 1.5 10 *2.00/0.* 8cc~\i.isc---<.)r ·ils~·I)O-'·()lis---stl~ucl·ure, ca..Q~J~I.~\_ IOlls~~~=\_.I:..'~\_s a la-j";gc c~:.pa.::itS~\_ (61~ enei·gy..sto,~ge

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Standardized bone specimen in a testing machine\_ The strain in the segment of bone between the two gauge arms is measured with a strain gauge. The stress is calculated from the total load measured. *Courtesy of Dennis R. Carter. Ph.D.*

(Ke"""ny & Hayes, 1993). The physical difference betwecn the two bone tissucs is quantified in terms of the apparent density of bone, which is ddined\_as the mas~\_.<?Ip·9ii~jI~~u£:i?,:·~~~ntin ~~f11e (gram per cubic ccn-titnctci:Tglcc]):-Frg"Llre a unit of bon~ \'01- 2-9 depicts typical stress-strain qualities of cortical and trabecular bone with different bone densities tested under similar conditions. In general, it is not enough lo describe bone strength with a single number. A bctter way is to examine the stress-strain curve for the bone tissue under the circumstances tested.

To better understand th" relationship of bone 10 other materials, schematic stress-strain curves for bone, metal, and glass illustrate the differences in mechanical behavior among these matcrials (Fig. 2~10). The variations in stiffness are rcOecled in the different slopes of the cun'cs in the clastic region. Metal has the steepest slope and is thLis the stiffest material.

**•**

C' .••----..-•••--------••••........••••........••••....

CPlastiC region

B'A B" C" Stress-strain curve for a cortical bone sample tested in tension (pulled). Yield point (B): point past which some permanent deformation of the bone sample oc- curred. Yield stress (8'): load per unit area sustained by the bone sample before plastic deformation took place. Yield strain (8"): amount of deformation with~ stood by the sample before plastic deformation oc- curred. The strain at any point in the elastic region of the curve is proportional to the stress at that point. Ultimate failure point (C): the point past which failure of the sample occurred. Ultimate stress (C'): load per unit area sustained by the sample be~ fore failure. Ultimate strain (C"): amount of defor- mation sustained by the sample before failure.

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Apparent Densily

scnce of a plastic region on the stl"L'ss-strain curve. ~:::1 CortIcal . bone Strain (%)

..... *0.30* glee

By contrasl, metal exhibits cxtensh·t: deformation

- - . *0.90* glee

- - 1.85 *glee*

before failing, as indicalC'd b~' a long plastic region on the ClitYC. Bone also deforms before failing but to a rnuch lesser extent than metal. The difference ~ iii ~*100* 50/

-------------- o.j.........,=.:...:..:".+"..:..:..".:....:..:.:..:".:."-'<":..:.":..:..:.. o

/ 15

in the plastic behavior of metal and bone is the l"e- suit or differences in microlllcchanical events at yield. Yielding in melal (tested in tension, or pulled) is caused b~' plastic rIow and the fonrwtion

/

of plaslic slip lines; slip lines art: formed when the molecules of lhe latticc structurc of mctal dislo- 520 25 cate. Yielding in bone (tested in tcnsion) is caused Trabecular .:..:...:.c..c.;.,:..:.":..•.c.;":..:•.:.":...".f.:.c.;".:.":..".:.."---< bone 10 Metal Example of stress-strain curves of cortical and trabec- ular bone with different apparent densities, Testing was performed in compression. The figure depicts the difference in mechanical behavior for the two bone structures. *Reprinted with permission from Ke,1'.'eny. T*M., & *Hc1yes, \tV.* C. (1993). *Mechanical properri(;.ls of cor- tical ancl rfDoecular bone,* Bone. 7, *28S·]t]4,*

The clastic portion of the curve for glass und metal is a straight linc. indicating linearly ei<:\slic be- havior; virtually no yielding lakes place before the yield point is reached. By comparison. precise test-

Bone ing of cortical bone hns shown thnt the elastic por- tion of the curve is not straight but instead slightly curved. indicating that bone is not linearly claslic in its bdlm'ior but yields somewhat during loading in the elastic rcgion (Bonefield & Li, 1967)" Table 2-1 depicts the mechanical propcrtics of selectcd bio· materials for comparison. Materials are classified as brittle or ductile depcnding on the extent of defor· mation before failure. Glass is a typical brittle ma- terial. and soft metal is a typical ductile material. The cUrrerence in the amount or deformation is re- Oected in the fracture surfaces or the two materials (Fig" 2-11)" When pieced togethcr aftcr fracture, lhe ductile material will not conform to its original shape whet'cas the brittle material will. Bone ex· hibits more brittle or ITIOI'C ductile behavior dt> pending on its age (younger bone being more duc- tile) and the rate at which it is loaded (bonc bcing more brittle at higher loading speeds).

After the yield point is reached, glass deforms very little before failing, as indicated by the ab·

Glass

Strain

Schematic stress-strain curves for three materials. Metal has the steepest slope in the elastic region and is thus the stiffest material. The elastic portion of the curve for metal is a straight line, indicating linearly elastic behavior. The fact that metal has a long plas- tic region indicates that this typical ductile material deforms extensively before failure. Glass, a brittle material, exhibits linearly elastic behavior but fails abruptly with little deformation, as indicated by the lack of a plastic region on the stress-strain curve. Bone possesses both ductile and brittle qualities demonstrated by a slight curve in the elastic region, which indicates some yielding during loading within this region.

**•**

~I i Mechanical Properties Biomaterials *of* Selected I

Ultimate Strength (MPa)

Metals

Co-Cr alloy

Cast 600 220 forged 950 220 15 Stainless steel 850 210 10 Titanium 900 110 15

Polymers

Bone cement 20 2.0 2-4,.-; Ceramic

Alumina 300 350 <2

Biological

Cortical bone 100-150 10-15 1-3 Trabecular bone 8-50 2-4 Tendon, ligament 20-35 2.0-4,0 10-25

Adapted from Kummer, J,K, (1999) Impldn, biomillcrials In J.M. Spivak, P.E, DiCesare, Os. Fe[dman, K,!. Ko'''.1[, A.S. RokilO, & J.D. Zuckerrnan (Eds.). *Orrhopaedic5.' A Study GlII'de* (pp. 45-48). Ni:>,v York: lvlcGraw-Hlil.

iIIIIIBrittle fracture ~ ,urface' of sample, of a ductile and a br;ttle I material. The *broken Jines* on the ductile material in-

dicate the original length of the sample. before it deformed. The brittle material deformed very little before fracture.

Ductile fracture

-.

Scanning electron photomicrograph of a human cor- tical bone specimen tested in compression (30X). *Ar- rows* indicate oblique cracking of the osteons. *Courtesy 0; Dennis R. Carter, Ph.D.*

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Modulus (GPa)

Elongation' (%)

by dcbonding of thL' ostC'ons 4\t the cement lines and micl'ofracturc (Fig. 2-12), while ~'idding in bone as a result of compression is indicated by cracking of lhe osteons (Fig. 2-13).

Because the structure of bone is dissimilar in the transverse and longillidinal. directions, it exhibits diffcrerl'l mechanical propenics whc.I1 loaded along differenl axes, a characleristic known as anisotropy.

Reflected-light photomicrograph of a human cortical bone specimen tested in tension (30):). *Arrows* indi- cate debonding at the cement lines and pulling out *1*

of the osteoos. *Courtesy* 0; *Dennis R. Carter. Ph.D.* t--------

.;;~~iY~C?;;ART

*200* MPa

Strain

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Anisotropic behavior of cortical bone specimens from a human femoral shaft tested in tension (pulled) in four directions: longitudinal (L). tilted 30" with respect to

**•**

the neutral axis of the bone, tilted 60", and transverse (T). Data from Frankel, V.H., & Burstein, A.H. (1970). Or- lhopaedic Biomech<1nlcs. Philadelphia: Lea & Febiger. ,

f·7jgure 2-14 shows the variations in strength and stiffness for cortical bone samples from a human remoral shah, tested in tension in four directions (Frankel & Burstein c 1970; Carterc 1978). The values for both parameters are highest for the samples loaded in the longitudinal direction. Figures 2·9 and 2·15 show trabecular bone slI-cngth and stiffness tesled in two directions: compression and tension. Trabecular or cancellous bone is approximately 25% as dense, 5 to *look;* as stiff, and five times as ductile as conical bone.

Although lhe relationship belween loading pat- lerns and the mechanical properties of bone throughout the skeleton is extremely complex, it generally can be said that bone strength and stiff- ness are greatest in the dircction in which daily loads arc most commonly imposed.

*BiOlnechanical Behavior* of *Bone*

The mechanical behavior or bone-ils behavior under the innuence of forces and moments-is af- fected by its mechanical properties, its geometric characteristics, the loading mode applied, direc- lion of loading, rale of loading, and frequency of loading.

10B6420-1----1----1----1---+--"'"-1 o 2 4 6 B 10 Tensile strain (%)

Example of tensile stress-strain behavior of trabecu- lar bone tested in the longitudinal axial direction of the bone. *Adapted from Gibson, L.1.,* & *As/lby. M.F. (1988). Cellular Solids: Structure and Properties.* New York: Pergamon, Press.

pendicular to the applied load (Fig. 2-17). Under lel)- sil~l(),l(ling,the stnl.c:ture lengthens and narrows.

Clinically, fractures produced by tensilelqading are usually seen in Dones with a large proportion of .. TensionCompression cancellous bone. Examples are fractures of the base or thCfifth metatarsal adjacent to the attachment of the pe-roneus brevis tendon and fractt,lres of the calcaneus adjacent to the attachment of the AchUlqs tendon. Figure 2-18 shows a tensile fracture through the calcaneus; intense contraction of the

Bending

triceps surae muscle produces abnormall,Y high ten- sile loads on the bone. •

•

I I'~ . eli I C..J i ShearCombined loading

Schematic representation of various loading modes.

During tensile loading, equal and opposite loads are applied olltward from the surface of the structure, ancCtensile stress and strai"n result inside the struc- ture. Ten'sile stress can be thought of as manv small forces directed·<.nva:v from the-;'urface of th~ stlJIC- lure. Maximal tensile stress occurs on a plane per-

Compression

DUling compressive loading, equal ~nd opposite lOi:lds are applied toward the surface of the structureaI1d compressive stress and strain. result inside the stI~IC­ ture. Compressive stress can be thought of as manyl small forces directed into the surface of the stlucture. Maximal compressive stress occurs on a plane per- pendicular to the applied load (Fig. 2-19). Under com- Torsion

pressive loading, the structure shortens and widens. ~L.-.- 1

Clinically, COinpression fractures are commonly

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found in the vertebrae. which are subjected to high compressive loads. These fractures are most often seen in the elderly with osteoporotic bone tissue. Figure 2-20 shows the shortening and widening

BONE BEHAVIOR UNDER VARIOUS LOADING MODES

Forces and moments can be applied to a structure in various directions, producing tension, compres- sion, bending, shear, torsion, and cornbinecl loading (Fig. 2-16). Bone in vivo is subjected to all of these loading modes. The folknving descriptions of these modes apply to structures in equilibrium (at rest or moving at a constant speed); loading produces an internal, deforming efFect on the structure.

Tension

Tensile loading.

Tensile fracture through the calcaneus produced by strong contraction of the triceps surae muscle during a tennis match. *Courtesy of Robert A. Winquisr, lA.D*

that takes place in a human vertebra subjected to a high compressive load. **In** a joint, compressive load- ing to failure can be produced by abnormally strong contraction of the surrounding muscles. An example of this effect is presented in Figure 2-2 t; bilateral subcapilal fractures of the Femoral neck

Compression fracture of a human first lumbar verte- bra. The vertebra has shortened and widened.

**•**were sustained by a patient undergoing electrocon- vulsive therapy; strong contractions of the muscles around the hip joint compressed the femoral head against the acetabulum. ..

Shear

During shear loading, a load is applied parallel to the surface of the structure, and shear,stress and strain result inside the structure. Shear stress can be thought of as many' small forces acting on the sur- face of the structure on a plane parallel to the ap- plied load (Fig. 2-22). *J\* structure subjected to a Compressive loading.shear load deforms internaH.v in an angular-rllanner; right angles on a plane surface within the stnlcture

*f*;~

1Dfm \_ eII.F- I 1

Bilateral subcapital compression fractures of the .------------------ femoral neck in a patient who underwent electrocon· vulsive therapy.

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-~

daslicily (Young's modulus) is approximately 17 GPa in longitudinal or axial loading and approxi- 1l1alcly I I GPa in transverse loading. Human tra- become obtuse or acute (Fig. 2-23'). \'Vhel~e\'er a structure is subjected to tensile or compressivelqad-

becular bone values for testing in compression are approximately' 50 ivlpa and arc reduced lO approxi- ing, sflear stl:ess is produced. -Figli;·c 2-24 iIllls11:a'tcs angtiliii· deformatiorl in sfi~lctllrcs subjected (0 these loading modes. Clinically, sh<;ar fractures are most ohen seen in cancellOlls bone.

Human «clllit 'corticai bone exhibits different val- ues for ultimate stress uncleI' compressive, tensile, and shear loading. Cortical bone can withstand greater stress in compression (approximately 190 Mpa) than in tension (approximately 130 !\Ilpa) and greater stress in tension than in shear (70 r\'lpa). The

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\_ Shear loading.

Force LBefore loading

When a structure is loaded in shear, lines originally at right angles on a plane surface within the struc- ture change their orientation, and the angle becomes obtuse or acute. This angular deformation indicates shear strain.

<> DUnloadedThe presence of shear strain in a structure loaded in tension and in compres.sion is indicated by angular deformation.

Under shear loading

**\*\*\*** o01

Under lensile loading<> CJ**\*\*\***

Under compressive loading

Cross-section of a bone subjected to bending, show- ing distribution of stresses around the neutral axis. Tensile stresses act on the superior side, and compres- sive stresses act on the inferior side. The stresses are highest at the periphery of the bone and lowest near the neutral axis. The tensile and compressive stresses are unequal because the bone is asymmetrical.

mately 8 Mpa iF loaded in tension. The modulus or elasticity is low (0.0-0.4 GPa) and dependent on the apparent density of the trabecular bone and direc~ lion of loading. The clinical biomechanical conse- quence is that the direction of compression failure results in general in a stable fracture, while a frac- ture initiated by..' tension or shear ma!' have cata- strophic consequences,

.... **..** 1\_/ 1\_/\_ \_B\_-----.JO A\_-------.JO .... **..**

Two types of bending. A, Three-point bending. B, Four-point bending.

lateral roentgenogram of a "boot top" fracture pro- duced by three-point bending. *Courtesy of Robert A \J1/inquist,* M, *D.*

Bending

In bending, loads are applied to a structure in a manner that causes it to bend ~\boutan,axis. \Vhen a bone 'is loaded in bending, it is subjected to a combination of tension and compression. Tensile stresses and strains act on one side of the neutral axis!.\_'111d compressive stresses and strain?,Ic;t on the oth~r side (Fig. 2-25); there are I}ostresse,:-;"lnd str~liI!?alongtt;enelltral axis. The ~lagnitlldeof the stresses is proportional to their distance from the neutral axis of the bone. The farther the stresses are from the neutral axis, the higher their magnitude. Because a bone structure is asymmetrical, the stresses may not be equally distributed.

Bending may be produced by thl'qe forces (three- point bendiilg) Ol:.Xour forces (roUI'~point bending) (Fig. 2-26). Fractures produced b)' both types of bending are commonly observed clinically, partiClI- lady in the long bones.

Three-point bendirlgtakesplace when three forces acting on a structure pn)c!uce two equal mo- mcnts, each being the product of one of the t\\'o pc- ripfle!~;;liforces and its perpcndicular distance. from the axis of rotation (the point at which the middle Forceis applied) (Fig. 2-26;\). IF loading continues to the )-'iele! point, the structure, if homogeneous, symmetrical, and with no structural or tissue de-

~ ,"'''' o Fatigued muscle ~

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feet, will break at the point of application of the middle force.

A typict;~Jthree~pointl)eIl(ling fracture is tht: "boot top"Jracturesustainedb:vskiers. In the "boot top" fracture shown in Figure 2-27, one bending moment acted on the proximal tibia as the skier fell forward over the top of the ski boot. An equal moment, pro- duced by the fixed foot and ski, acted on thc distal tibia. As the proximal tibia was bent fonvard, tensile stresses and strains acted on the posterior side of the bone and compressive stresses and strains acted on the anterior side. The tibia and fibula fractured at the top of the boot. Because adult bone is weaker in tension than in compression, failure begins on the side subjected to tension. Because immature hone is n10r~ ductile, it nUl)' fail first in compres- sion, and a buckle fracture may result on the com- pressive side (Flowchart 2~ I).

Four~point bending takes place when two force couples acting on a stl'uclurc produce two equal moments. A force couple is formed when two paral~ lei forces or equal magnitude but opposite direction are applied to a structure (Fig. 2-28;\). Because the magnitude of the bending moment is the same throughout the area between the two force couples, the structure breaks at its weakest point. An exam- ple of a FourNpoint bending FraclUre is shown in Fig~ ure *2-28B.* AstifT knee joint was manipulated incor- rectly' during rehabilitation of a palient with a postsurgical infected femoral fracture, During the

• A

A, During manipulation of a stiff knee joint during fracture rehabilitation, four-point bending caused the femur to refracture at its weakest point, the original fracture site. B, Lateral radiograph of the fractured femur. *Courtesy of Kaj Lundborg,* M.D.

Cross-section of a cylinder loaded in torsion, showing the distribution of shear stresses around the neutral axis. The magnitude of the stresses is highest at the periphery of the cylinder and lowest near the neutral axis. Experimentally produced torsional fracture of a ca·

nine femur. The short crack *(arrow)* parallel to the neutral axis represents shear failure; the fracture line at a 30'" angle to the neutral axis represents the plane of maximal tensile stress. manipulation. the posteJ"ior knee joint capsule and tibia Formed one force couple and the fClTlOral head and hip joint capsule formed the other. As a bending moment was applied to the femur, the bone failed at its \vcakest point, the original fracture site.

Torsion

In torsion, a load is applie~l to a strl1c~.lII·c \_in a man- nel' that causes i"l to l\\;j'st about an axis, and a lO~'gue--(or ornament) is produced within the-struc- lure.--\Vhen a structure is loaded in torsion, shear

Combined loading,

Allhough each loading mode has been considered separately, living bone is seldom loaded in one mode only. Loading of bone in vivo is complex for two ,;

principal reasons: bones .~.re ~onstantlysubjected to multiple indeterminate loads and their geomelric structure is irregulm: In vivo mcasurement of the strains on the antcrol11cdial surface of a human adull tibia during walking and jogging dcmon- stre:;ses arc distributed over the entire structure. As in b~n'ding, the magnitude of these stresses is p1'o- ponional to their distance from the neutral axis (Fig. 2·29). The farther the stresses arc from the neutral axis, the higher their magnitude.

Under torsional loading, maximal shear stresses act on plan~sp~~-;'alleland~perpen(ficular·to-theneu- tl"al axis of the structure. 'In addition, max"fillal ten- sile aild comp'ressi\'c stresses act on a plane diago- nal to the neutral a.'\is of the structure. Figure 2-30 illustrates these planes in a small segment of bone loaded in torsion. <J)61, , , , , . i I , '! '

~~@.- Shear

Tension

The fracture pattern for bone loaded in torsion suggests thal the bone fails first in shear, with the formation of an initial crack parallel to the neutral axis along of the the plane bone. or A maximal second tensile crack usually forms stress. Such a ,i<?

, : pattern can be seen in the experimentally produced torsional fracture of n canine femur shown in Fig- ure 2-31. **.......** '-.-'I **1** Compression Schematic representation of a small segment of bone loaded in torsion. Maximal shear stresses act on planes parallel and perpendicular to the neutral axis. Maximal tensile and compressive stresses act on planes diagonal to this axis.

SlrateS the comph.:.xil.V 01" the loading patterns dur- ing these cOl11mon ph.'·siological activities (Lanyon el aI., 1975). Stress values calculated from these strain measurel~lentsby Carter (1978) showed thal during normal walking, the strc.i.~es wcre COlllP.I"CS- s.ivc dllring heel strike, tensile during- thc--s\_t-~!.nce ph~y~·,·,indi1¥aill--c()il·~prcssive-.~Iu'ri,~,lg- pLlsl.1~off (Fig. *2-3"2A).* VaI"ucs for shear stress \\'cn.:.~ relatively high in the later portion of the gait c)ldc, denoting sig- nificant torsional loading. This torsional loading \vas associated with external rotation of the tibia during stance ancl push-ofr.

Dul"ing jogging. the stress pattern was quite dif- rerent (Fi-g. 2:328). The COlJlP.~·.~~.~~c\_slrcssPFcdom- inatin!! at lac strike was followed bv high tensile stress~di.iring r>ll-sh-ofr. The sllenr slrcss~:vi~~--'o\V tfil'oughout 'tlle strich:\ denoting111ini;lwl torsional loadhig"pl~o(lllcedby slight e~'\tern~" and inlcrm~l 1'0- tatiOll or-the tibia in an-ahernating pallcrn. The in- crease--fll specd from slow walking LO jogging in-

34 Walking (1.4 *m/sec)*

StressTensile

Compressive 2 Shear (external rotation) ~ 1 I z 1 ;;; 0+''<,--+-!----o:...-.--+,\_\_+ \_\_.-;~'c ...,...+- ~"' ;i •••••••• ' -', *..j ...i* {jj \. ! -~'.. .:; \ .....;, ... *i* \·······1··..//

\.. i "r 3 i 4· h. *l,\* HS FF A HO

A. Calculated stresses on the anterolateral cortex of a human tibia during walking. HS, heel strike; FF, foot flat; HO, heel-off; TO. toe off; S, swing. *Calwlated from Lanyon, L.E.. Hampson.* W'.G.J., *Goodship. A.E.. et af. (1975). Bone deformation recorded in vivo (rom srrain gauges c1tr<elched to the human tibial sharr.* ACla Orlhop Scand, 46. 256. *Figure coortesy of Dennis R. Carler, Ph.D*

s HOTO

crcas~d both the str~ss and the strain Oil the libia (Lan.'·on ct aI., 1975). This increase in strain with grt,,:atcr speed was confirmed in studies or locomo~ lion in sheep, which ckmonslratcd a fivefold in- crcase in slrain values from siD\\" walking to fast lrotting (Lanyon & Bourn, 1979).

INFLUENCE OF MUSCLE ACTIVITY ON STRESS DISTRIBUTION IN BONE

When bone is loaded in vivo, the contraction of tbe muscles attached to the bone alters dlC stress distri- bution in [he bone. This muscle contraction de- cr<.:ases 01· eliminates [ensile stress on the bone by producing compressi\'c SlI-CSS thal neutralizes it ei- ther partially or totall~·.

The effect of muscle contraction can be illus~ trated in a tibia subjected to three-point bending. Figure 2~33A represents Lhe leg of a skier who is falling: forward, subjecting the tibia Lo a bending

12Jogging (2.2 *m/sec)*

10Tensile

8

Compressive Shear (external rotation) Shear (internal rotation) rn a.6 ~~ 4 ~iii ~O' , i

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"' "' "' \ *\i* i i i i B 2

2·

*4* .l---=-E--=~'-:':::------ TS *1\* TS·TO

8, Calculated stresses on the anterolateral cortex of a human tibia during jogging. T5, toe strike; TO, toe off. *Calwfared [rom Lanyon. L.E.. Hampson.* W.G.)., *Goodship.* AE., *et al.* (1975). *Bone deformation recorded in vivo from strain gaoges aClacIJed* ro *the human tibial Shaft.* Acta Orthop 5cand. 46. 256. *Figure courtesy 01 Dennis R. Carter, Pll.D.*

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A. Distribution of compressive and tensile stresses in a tibia subjected to three-point bending. B, Contrac- tion of the triceps surae muscle produces high com- pressive stress on the posterior aspect, neutralizing the high tensile stress.

moment. High tensile stress is produced on the pos- tcrior aspect of the tibia, and high compressive stress acts on the anterior aspect. Contraction of the triceps surae muscle produces great compressive stress on the posterior aspect (Fig. 2-338), neutral- izing the great tensile stress and thereby' protecting the tibia from failure in tension. This muscle con- traction may result in higher compressive stress on the anterior surface of the tibia and thus protect the bone from failure. Adult bone can usuall~/withstand this stress, but immature bone, which is weaker, ma.y fail in compression.

Muscle contraction produces a similar effect in the hip joint (Fig. 2-34). During locomotion, bend- ing moments are applied to the femoral neck and tensile stress is produced on the superior cortex. Contraction of the gluteus medius muscle pro- duces compressive stress that neutralizes this ten- sile stress, with the net result that neither com- pressive nor tensile stress acts on the superior cortex. Thus, the muscle contraction allows the femoral neck to sustain higher loads than would otherwise be possible.

STRAIN RATE DEPENDENCY IN BONE

Because bone is a viscoelastic material, its biome- chanical behavior varies with the rate at which the

bone is loaded (Le., the rate at which the load is ap- plied and removed). Bone is stifTer and sustains a higher load to failure when loads are applied at higher rates. Bone also stores more energy before failure at higher loading rates, provided that these rates are within the physiological range.

The in vivo claily' strain can vary considerably.'. The calculated strain rate for slow walking is 0.001 per second, \vhile slow running displays a strain rate of 0.03 per second.

In general, when activities become more strenu- ous, the strain rate increases (Keaveny & Hay'es, 1993). Figure 2-35 shows cortical bone behavior in tensile testing at different physiological strain rates. As can be seen from the figure, the same change in strain rate produces a larger change in ultimate stress (strength) than in elasticit.\' (Young's modu- lus). The data indicates that the bone is approxi- mately *30(/0* stronger for brisk walking than for slow

"*E,,--,,-,* , ' , ' , ' , ' ; : ,,c"~ ,/ ,~ : : , ,, ,

Stress distribution in a femoral neck subjected to bending. When the gluteus medius muscle is relaxed *(top),* tensile stress acts on the superior cortex and compressive stress acts on the inferior cortex. Con- traction of this muscle *(bottom)* neutralizes the ten- sile stress.

400

walking. At very high strain rates (> I per second) representing impact trauma, the bone becomes more brittle. In a full range of cxperimentaltesting for ultimate tensile strength and elasticity of corti- , cal bone, the strength increases by a factor of three and the modulus by a faclor of two (Keaveny & Hayes, 1993).

The loading rale is clinically significant because it innuences both the fracture paUern and the amount of soft tissue damage at fracture. \Vhen a bone frac- tures. the stored energy is released. At a low loading rate, the energy can dissipate through the formation of a single crack; the bone and soft tissues remain relatively intact, with little or no displacement of the ,. bone fragments. At a high loading rate, however, the ;'·',greater energy stored cannOt dissipate rapidly :~ enough through a single crack, and comminution of bone and extensive soft tissue damage resull. Figure . 2-36 shows a human tibia tested in vitro in torsion at a high loading 1'4ue; numerou:j bone fragments *;;J'*

1500/sec 300

300/sec

I~ l::l "200 'I"iiiO.OOl/sec

Slow walking

o'l'---+---+---+------i 0.0 0.5 Rate dependency of cortical bone is demonstrated at five strain rates. 80th stiffness (modulus) and strength increase considerably at increased strain rates. *Adapted from McElhaney, J.H.* (1966). *Dynamic* I,*response of bone and muscle tissue.* J Appl Physio1, 2},

/23/-/236.

III

~--- O.lIsec - - - - - O.Ollsec

Brisk walking

100

1.5

2.0

were produced, and displacement of the fragments was pronounced.

Clinically, bone fractures fall into three general categories based on the amount of energy released at fracture: low-energy!, high-energy, and VCI)' high- energy. A low-energy fracture is exemplifled by the simple torsional ski fracture; a high-energy fracture is often sustained during automobile accidents; and a very high-energy fracture is produced by vcry high-muzzle velocity gunshot.

FATIGUE OF BONE UNDER REPETITIVE LOADING

Bone fractures can be produced b:v a single load that exceeds the ultimate strength of the bone or by repeated applications of a load of lower magni- tude. A fracture caused by a repeated load applica- tion is called a fatigue fracture and is lypically pro- duced either by few repetitions of a high load or by many repetitions of a relatively normal load (Case Study 2-1).

The interplay of load and repetition for any ma- te!"inl can be plotted on a fatigue curve (Fig. 2-37), For some materials (some metals. for example), thc fatiguc curve is asymptotic, indicating that if the load is kept below a certain level, theoretically the material will remain intact no matter how many repetitions. For bone tested in vitro, the curve is not asymptotic. \,Vhen bone is subjected to repetitive low loads, it may sustain microfractures. Testing of bone in vitro also reveals that bone fa- proaches tigues rapidly its yield when strength; the load that or is, deformation the number ap- or repetitions needed to produce a Fracture dimin- ishes rapidly.

In repetitive loading of living bone, the fatigue process is affected not only by the amount of load and the number of repetitions but also bv the number of applications of the load within a given time (frequency of loading). Because living bone is self-repairing. a fatigue fracture results only when the remodeling process is outpaced by the fatigue process-that is, when loading is so Frequent that it precludes the remodeling necessary to prevent failure.

Fatigue fractures are llsually sustained during continuous strenuous physical activity, which causes the muscles to become fatigued and reduces .' their ability to contracl. As a result, the.v are less able to store energy aI~d thus to neutralize the stresses imposed on the bone, The resulling alter- ation of the stress distribution in the bone causes

Human tibia experimentally tested to failure in torsion at a high loading rate. Dis- placement of the numerous fragments was pronounced.

B"ne Overloading A" , 23-year-old military recruit was exposed to an intensive

' heavy physical training regime that included repetitive continuous crawling in an awkward position for several weeks (Case Study Fig. 2-1-1 A). The repeated application of loads (high repetitions) and the number of applications of a load during a short period of time (high frequency of loading) surpassed the time for the bone remodeling process to pre-

vent failure. iVluscie fatigue occurred as a result of the abnor- mal loading pattern and the intensive uaining, It affected the muscle function in the neutralization of the stress imposed, leading to abnormal loading and altered stress distribution (Case Study Fig. 2-1-1 B).

After 4 \-veeks of strenuous physical activity, the damage accumulation from fatigue at the femoral shaft lead to an oblique fracture.

Case Study Figure 2-1-1A. Abnormal loads at the femoral shaft occurred.

duced more slowl~'; the remodeling is Icss casily out- paced by the fatiguc process and the bone may not proceed to complete fracture. ,'.This theol)1 of muscle fatigue as a cau~e of fa- tigue fracture in the lower extremities is outlined in the schema in Flowchart 2-1 on p. 41.

Figure 2-38 shows typical strain ranges for hu- man femoral cortical bone during different activi- tics and distances. Resistance to fatiguc behavior is great.er in compression than in tension (Keaveny & Hayes, 1993). On average, approximately 5,000 cy- clcs number of experimental of steps in to loading miles of correspond running. One to mil- the

lion cycles corresponds to approximately 1,000

•miles. A total distance of less than 1,000 miles could cause a fracture of the cortical bone tissue. This is consistent with stress fractures reported among military recruits undergoing strenuous training of up *to* 1,000 miles of nll1ning over a short period of timc (6 weeks). Fracturcs of indi- vidual trabeculae in cancellous bone have been ob- served in postmortem hUlllan specimens and Illay be caused by fatigue accumulation. Common sites arc the lumbar vcnebrae, the femoral head, and II::1 I~ ·tI) II i 'E l~g ~ the proximal tibia. It has been suggested that these fractures may playa role in bone remodeling as well as in age-related fractures, collapse of sub- chondral bone, degeneraLive joint diseases, and other bone disorders.

0006 0 004 vlgorous,-:":::!::::""~~S~'b-'<:::::::-;:;-~ 0.002 Running exercise -----------===--:::-

Walking

"i

".3 mThe interplay of load and repetition is represented on a fatigue curve.

Injury

Repetition

abnormally high loads to be imposed, and n fatigue damage accumulation occurs that Illav lead to a fracture. Bone may fail on the tensil~ side, on the compressive side, or on both sides. Failure on the tensile side resulls in a transverse crack, and the bone proceeds rapidly to complete fracture. Fatigue fnlctures on the compressive side appear to be pro-

o Compression

• TenSion

Miles 10• I100•

I1000!

INFLUENCE OF BONE GEOMETRY ON BIOMECHANICAL BEHAVIOR

The geometrv of a bone greatl\' influences its mc-

0.0001+0-0--1---+---+----+---!------I 1~ 1~ Number 01 Cycles lam I --------

chanical bel~avior. In Le~lsion' and compression.

1~

the load to failure and the stiffness arc propor- tional to the cross-sectional area of the bone. The larger the area, the stronger and stiffer the bone. In bending, both the cross-sectional arca and the distribution of bone tissue around a neutral axis Fatigue testing showing the number of cycles (x-axis) and strain range (y-axis) expressed as stress rangel

affcct the bone's mechanical behavior. The quan- tity that takes inLO account these two factors in modulus in human cortical bone specimens loaded in tension and compression. Typical strain ranges are

bending is called the area moment of inertia. A larger moment of inertia results in a stronge.· . II and stiffer bone. Figure 2-39 shows the influ- ence of the arc,) moment of inenia on the load to failure and the stiffness of three rectangular Slruclures thal have the same area but different shapes. In bending, beam III is thc stillest of the lhree and call withstand the highest load because the greatest amount of material is distribuLed at a distance from the neulral axis. For rectangular cross-sections, the formula for the area momcnt of shown for walking, running. and vigorous exercises. Note that resistance to fatigue fracture is greater in compressive loading. Ten miles represent approxj· mately *5,000* cycles, corresponding to the number of I ' steps running during that distance. *Aclapared from Carrer. D.R., Cater. W.E., Spengler, O.M.. Frankel, \l.H.* ,i*(J* 98 1). *Fatigue behavior of adtilt cortical bone: the influ-*

*J.-----------------* i i *ence of mean srrc1ilJ anel strain range.* Acra Orthop Seane!. 52.48/-490.

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4 ){ 1

2x2

1 x 4 I

II

III

Three beams of equal area but different shapes subjected to bending. The area moment of inertia for beam I is 4/12; for beam II, *16112;* and for beam *111,64/12. Adapted hom Franke', VH..* & *Burstein, AH. (970).* Orthopaedic Bio- mechanics. *Philadelphia: Lea* & *Febiger.*

inertia is the width (8) multiplied by the cube of

The factors that affect bone strength and stiffness the height (1-1') divided by 12:

in torsion are the same that operate in bending: the

B· H' 12

cross-sectional area and the distribution of bone tis- sue around a ncutral axis. The quantity that takes into account these two factol's In torsional loading Because of its large area moment of inertia. bean"'!

is the polar moment of inertia. The larger the polar III can withstand four times more load in bending

moment of inertia. the stronger LInd stiffer the bone. limn can beam I.

Figure 2-41 shows distal LInd pl'oximnl cross- A third factor, the length of the bone, influences

sections of a tibia subjected to torsional loading. Al- the strength and stillness in bending. The longer the

though the proximal section has a slightly smaller bone, the greater the magnitude of the bending mo-

bony area thun docs the distal section, it has a much ment caused by the application of a force. ,In a rec-

higher polar moment of inertia because much of the tangular structure, the magniwde of the stresses

bone tissue is distributed at a distance from the neu- produced al the point of application of Ihe bending

tral axis. The distal section, while it has a largel" moment is proponional to lhe length of the StI1.IC-

bony area. is subjected to much higher shear stress lure. Figure 2-40 depicts the forces acting on two

bccause much of the bone tissue is distributed close beams with the same width and height but different

to the neutral axis. The magnitude of the shear lengths: beam B is twice as long as beam A. The

stress in the distal section is approximately double bending moment for the longer beatn is twice that

that in the proximal section. Clinically, torsional for the shorter beam; consequently, the stress mag~

fractures of the tibia commonly occur distnlly. nitudc throughout the beam is twice as high. Be-

When bone begills to heal after fracture, blood cause of their length, the long bones of the skeleton

vessels and connective tissue from the periosteum are subjected to high bending moments and. there-

migrate into the region of the fracture. forming a fore, to high tensile and c01npressive stresses. Their

cuff of dense fibrous lissue, or callus (woven bone), tubular shape gives them the ability to resist bend-

around the fracture site. stabilizing that area (Fig. ing moments in all directions. These bones have a

*2-42A).* The callus significantly increases the area large area moment of inertia because much of the

and polar moments of inertia. thereby increasing bone tissue is distributed at a distance from the neu-

the strength and stiffness of the bone in bending tral axis.

and torsion during the healing period. As the rrac~ l

Stress ~-"'~~-<:;magnilude S

*-r-\_*

I-L---... -L • I as as bending moment. Hence, the stress magnitude throughout beam B is twice as high. *Adapted from*

*VH.,* & *Burstein, A.H. (7970),* Orthopaedic Bio· mechanics. *Philadelphia: Lea* & *Febiger.*

Stress magnitude =, 25

2L \_\_\_\_\_

------ 2L ----.

I A, Early callus formation in a femoral fracture fixed with an intramedullary nail. B, Nine months after in- jury, the fracture has healed and most of the callus cuff has been resorbed. *Courtesy of Robert A. vVinquist, MD*

,,,,,,,,, ,,,,,,,,,~l.,;\_

ture heals and the bone gradually regains its normal strength, the callus cuff is progressively resorbed and the bone returns to as ncar its normal size and shape as possible (Fig. *2-4213).*

Certain surgical procedures produce defects that greatly weaken the bone, particularly in torsion. These defects fall into two categories: those whose length is less than the diameter of the bone (stress raisers) and those whose length exceeds the bone di- ameter (open section defects).

Astress raiser is produced surgically' when a small piece of bone is removed or a screw is inserted. Bone strength is reduced because the stresses imposed

Distribution of shear stress in two cross-sections of a tibia subjected to torsional loading. The proximal

during loading are prevented From being distributed evenly throughout the bone and instead become con- section (A) has a higher moment of inertia than does

centrated around the defect. This defect is analogous the distal section (B) because more bony material is

to a rock in a stream, which diverts the watel~ pro- distributed away from the neutral axis. *Adap(ed from*

ducing high water turbulence around it. The weak- *Franke!, VH.,* & *Burstein, AH, (1970;'* Orthopaedic Bio·

ening effect of a stress rqiser is particularly marked mechanics. *Philadelphia: Lea g Febiger.*

under torsional loading; the total decrease in bone strength in this loading mode can reach 60~o.

O},[~fN£'~T:l''. BIOMECHANICS OF TISSUES AND STRUCTURES Of THE MUSCULOSKELETAL SYSTEM

c-::-<........) Empty screw hole

raiser effect produced by the screws and by the 125

\_ -+-... Screw in place

hol('s without screws had disappeared completely \_:.~rew removed ~l\_te~..- 100 ~,2 [;;because the bone had ren1odclcd: bone had been laid down around the screws to stabilize them, and the empty screw holes had been filled in with bone. In femora from which the screws had heen re- moved immediately before testing, however, the

w 0> <;;~c75 *.,..---'y* ~-- . *.*

<I"

/' / \_.::....--- 50 energy storage capacity of the bone decreased b)' *500hi,* mainly because the bone tissue around the 25

screw sustained microdamage during sere\\' re- moval (Fig, 2-43),

a

2

345

6 7 B An open seclion defect is a discontinuity in the bone caused by the surgical removal of a piece of Weeks

bone longer than the bone's diameter (e.g.• by the clilling of a slot during a bone biopsy). Because the ollter surface of the bonc's cross-section is no longer

Effect of ~crews and of empty screw holes on the en- ergy storage capacity of rabbit femora. The energy storage for experimental animals is expressed as a

continuous. its abilit!1 to resist loads is allered, par- ticularly in torsion.

In a normal bone subjected to tOl'sion, the shear percentage of the total energy storage capacity for

stress is distributed throughout the bone and acts La control animals. When screws were removed immedi-

resist the torque. This stress pattcl-n is illustrated in ately before testing, the energy storage capacity de 4

the cross-section of a long bone shown in Figure creased by 50%. *Adapted from Bur51!.>in. A.H., et* al.

*2-44A.* (A cross-section with a continuous Ollter sur- *(1972), Bone strength: The effect of sere"v floles.* J Bone Joint Surg. 54A, ] I*i13.*

face is called a closed section.) In a bone with an

Burstein and associates (1972) showed the effect

Control

of stress rabel's produced by' screws and by empty screw holes on the energy storage capacity of rab- bit bones tested in torsion at a high loading rate. The irnmediatc effect or drilling a hole and insert- ing a screw in a rabbit femur was a *74%* decrease in energy stOl'age capacily. After 8 weeks, the stress

Open section

Load-deformation curves for human adult tibiae Stress pattern in an open and closed section under

tested in vitro under torsional loading. The control torsional loading. A, In the closed section, all the

curve represents a tibia with no defect; the open sec- shear stress resists the applied torque. B, In the open

tion curve represents a tibia with an open section de- section, only the shear stress at the periphery of the

fect. *Adapted from* Fr(lf)k~/. *v.H.,* & *Burstein, A.H. (1970).* bone resists the applied torque.

Orthopaedic BiomechaniCS. *Philadelphia: Leel* & *Febiger.* Deformation

graft was removed for use in an arthrodesis of the hip. A fe\v weeks after operation, the patient tripped while twisting and the bone fractured through the defect.

*Bone Remodeling*

Bone has the ability to remodel, b)' altering its size, shape, and structure, to meet the mechanical de- mands placed on it (Buckwalter et aI., 1995). This phenomenon, in which bone gains or loses cancel- lous and/or cortical bone in response to the level of stress sustained, is summarized as \'VqJJr's law, which states that the, remodeling of bone is influ- enced and modulated by mechanical stresses (Wolff, 1892): .

Load on the~,keleton can b~,,,~lc.l.:(?I""l~plishedby ei- tht:;,I.:,IX1~~\_?(.:lc,activily or gravity. A positive correlation exists b~twcen bOI1e mass and body we-ight. A gr~ater body weight has been associated \vitha larger bone mass (Exner et al., 1979). Conversely, a prolonged condition ohveightlcssness, such as that expedenced during space travel, has been found to result in de- creased bone mass in weight-bearing bones. Astro- nauts experience a fast loss of calcium and con- sequent bone loss (Rambaut & Johnston, 1979; \Vhedon, 1984). These changes are not completely reversible.

~'-------

Normal

..----------------- IIIA patient sustained a tibial fracture through a surgi- cally produced open section defect when she tripped a few weeks after the biopsy.

open section defect, only the shear stress at the pe- riphery of the bone resists the applied torque. As the shear stress encounters the discontinuity, it is forced to change direction (Fig. *2-44B),* Throughout the interior of the bone, the stress nms parallel to

Deformation the applied torque, and the amount of bone tissue resisting the load is greatly decreased.

In torsion tests in vitro of human adult tibiae, an open section defect reduced the load to failure and

Load-deformation curves for vertebral segments L5 to L7 from normal and immobilized Rhesus monkeys. energy storage to failure h,v as much as *90'10.* The de-

Note the extensive loss of strength and stiffness in formation to failure was diminished by! approxi-

the immobilized specimens. *Adapted from Kazarian,* mately 70% (Frankel & Burstein, 1970) (Fig. 2-45).

*L.L..* g *Von Gierke, H.E. (1969) Bone loss as* a *result of* Clinically, the surgical removal of a piece of bone

*immobilization and chelation, Preliminary results* in can greatly \veaken the bone, particularly in torsion. *ivlacaca mulatta.* (lin Orthop. 65. 67.Figure *2A6* is a radiograph of a tibia from which a

Bone Remodeling A30-year-old man who had a surgical removal of an .. ulna plate after stabilization of a displaced ulnar fracture, Figure 2-48 shows anteroposterior (A) and lateral (8) roentgenograms of the ulna after late plate removal. The implant is used to stabilize the fracture for rapid healing. However, in situations such as this, the late plate removal decreased the amount of mechanical stresses necessary for bone remodeling. It is of concern when the plate carries most or all of the mechanical load and re· mains after fracture healing. Thus, according to Wolff's law, it will promote localized osseous resorption as a re- sult of decreased mechanical stress and stimulus of the bone under the plate, resulting in a decrease in strength and stiffness of the bone.

Disuse or inactivity has deleterious effects on the skeleton. Bcd rest induces a bone mass de- crease of approximately' I*(Ve-* per week (Jenkins & Cochran, 1969; Krolner & Toft, 1983). In partial or total immobilization, bone is not subjected to the usual mechanical stresses, which leads to resorp-

Anteroposterior (A) and lateral (B) roentgenograms of an ulna after plate removal show a decreased bone diameter caused by resorption of the bone un- der the plate. Cancellization of the cortex and the presence of screw holes also weaken the bone. *Cour- tesy of Marc Marrens,* 1\.1. *D.*

Roentgenogram of a fractured femoral neck to which a nail plate was applied. loads are transmitted from the plate to the bone via the screws. Bone has been laid down around the screws to bear these loads.

tion of the periosteal and subperiosteal bone and a decrease in the mechanical properties of bone (Le., strength and stiffness). This decrease in bone strength and stiffness was sho\vn b:v Kazarian and Von Gierke (1969), who immobilized Rhesus mon- keys in full-body casts for 60 days. Subsequent compressive testing in vitro of the vertebrae from the immobilized monke.\!s and from controls showed up to a threefold decrease in load to failure and energy storage capacity in the vertebrae that had been immobilized; stiffness was also signifi- cantly decreased (Fig. 2-47).

An implant that remains firmly" attached to a bone after a fracture has healed may also diminish the strength and stiFfness o( the bone. **In** the case of a plate fixed to the bone with screws, the plate and the bone share the load in proportions deter-

c~•. ....'------------------- UW! nl nnc::7 UU.· o0 II ,'- ." Vertebral cross-sections from autopsy specimens of *(top)* is subjected to absorption *(shaded area)* during

,I

young (A) and old (8) bone show a marked reduction in the aging process, the longitudinal trabeculae become cancellous bone in the latter. *Reprinted with permission* thinner and some transverse trabeculae disappear *(bot-* l\_\_W\_i'\_h\_a\_I *from Nordin. B,E.e.* (1973). Metabolic Bone and Stone Dis- ease. *Edinburgh: Churchill Livingstone.* C. Bone reduction 9\_i\_n\_9\_i'\_'\_'\_h\_e\_m\_a\_t\_i,\_a\_'\_'Y\_d\_e\_P\_i'\_t\_e\_d\_.\_A\_,\_n\_o\_,\_m\_a\_1\_b\_o\_n\_e *tom). Adapted from Sifter!, R.S..* & *Levy. R.N. (J98/j. Tra- becular pacteffls and the internal architecture of bone.* ivlt. S\_i\_"\_"\_i\_J\_rv\_lo\_d\_,\_";\_.S\_,\_2\_2\_'\_. \_

mined by the geometry and n1mcrial properties of each structure (Case Stud)' 2-2). A large plate, car- rying high loads, unloads the bone to a great ex- tent; the bone then atrophies in response to this di- minished load, (The bone may hypertrophy at the bone-screw interface in an altcmpt to reduce the rnicrol1lotion of the screws.)

Bone resorption under a plate is illustrated in Figure 2-48. A compression plate made of a mater- ial approximately 10 times stifTer than the bone \Vas applied to a fractured ulna and remained after the fracture had healed, The bone under the plate ear- ried a lower land than normal; it was partially re- sorbed, and the diameter of the diaphysis became markedly smaller. A reduction in the size of the bone diameter greatly decreases bone strength. par- ticularly in bending and torsion, as it reduces the area and polar moments of inertia. A *20(M,* decrease in bone diameter may reduce the strength in tor- sion by *60%.* Changes in bone size and shape illus- trated in Figure 2-48 suggest that rigid plates should be removed shortly after a fraeture has healed and before the bone has markedly dimin- ished in size. Such a decrea.sc in bone size is usu- ally accompanied by secondary osteoporosis, which' further weakens the bone (SI'itis et a!., 1980),

An implant may cause bone hypertrophy at its at- tachment sites. An example of bone hypertrophy

around scrcws is illustrated in Figure 2-49. A nail plate was applied to a femoral neck fracture and the bone hypcrtrophied around the scrcws in responsc to the increased load at thesc sites. H.~lpertrophy may also result if bone is repeatcdly subjected to high mechanical stresses within the normal physio- logical range. Hypertrophy of normal adult bone in response to strenuous exercise has been observed (Dalen & Olsson, 1974; Huddleston et aI., 1980; Jones et aI., 1977), as has an increase in bone den- sity (Nilsson & Wesllin, 1971),

*Degenerative Changes in Bone Associated With Aging*

A progressive loss of bone density has been ob- sCI·veci as part of the normal aging process. The lon- gitudinal trabeculae become thinner, and some of the transverse trabeculae arc resorbed (SifTert & Levy, 1981) (Fig. 2-50). The result is a marked re- duction in the amount of cancellous bone and a thinning of conical bone. The relationship between bone mass, age. and gender is shown in Figure 2-51. The in the decrease size or ill the bone bone Lissuc reduce and bone the slight strength decrease and stillness.

or

20 40 60 BO Age (years) ~OWlng the relatIOnship between bone mass,

age, and gender. On the top of the figure, a cross- section of the diaphysis of the femur and the bone mass configuration is shown. *Reprinted ('lith permission from Kaplan, F.S., rlayes, W.c., Keaveny, T.M.,* er *al (7994), Form and function of bone. In S.R. Simon* (edJ Orthopaedic Basic Science *(.0,* 767). *Rosemont, IL:AAOS*

Stress~strain curves for specimens from human adult tibiae of t\VO widely differing ages tested in Uoro:;; E'"*ww*mIDtIl Co1000

tension are shown in Figure 2~52. The ultimate stress was approximately the sarne for the .young and the old bone. The old bone specimen could

500

withstand only half the strain that the young bone could, indicating greater brittleness and a reduction in energy storage capacit},. The reduction in bone density', strength, and stiffness results in increased bone fragility!. Age~related bone loss depends on a number of factors, including genclel: age, post- menopause, endocrine abnormality, inactivity, dis- use, and calcium deficiency. Over several decades, the skeletal mass ma.y be reduced to 5000 of original trabecular and *25°/(;* of cortical mass. In the fourth decade, women lose approximately 1.5 to *2(10* a year while men lose only approximately half that rate (0.5 to *0.75(-/0)* yearly. Regular physical activity and exercise (Zetterbarg et aI., 1990), calcium, and pos~ sibly estrogen intake may decrease the rate of bone mineral loss during aging.

*Summarv*

Strain

Stress-strain curves for samples of adult young and old human tibiae tested in tension, Note that the bone strength is comparable but that the old bone is more brittle and has lost its ability to deform. *Adapted from Bursrein, A.H., Reilly,* D. *T,* & *Alartens,* M. *(976). Aging of bone tissue' Mechanical properties.* Bone Joint Surg, 58A, 82.

*1i* Bone is a complex two-phase composite mate~ rial. One phase is composed of inorganic mineral salts and the other is an organic matrix of collagen and ground substance. The inorganic component makes bone hard and rigid, whereas the organic component gives bone its flexibility and resilience.

2 tVlicroscopically, the fundamental structural unit of bone is the osteon, or haversian system, composed of concentric layers of a mineralized ma~ trix surrounding a central canal containing blood vessels and nellie fibers.

3· rv(acroscopicall.\" the skeleton is composed of cortical and cancellous (trabecular) bone. Cortical bone has high density' while trabecular bone varies in density over a wide range.

Bone is an anisotropic material, exhibiting diF- ferent mechanical properties \vhen loaded in differ- ent directions. iVlature bone is strongest and stiffest in compression.

S Bone is subjected to complex loading patterns during common physiological activities such as walking and jogging. Most bone fractures are pro- duced by a con1bination of several loading modes.

6 Ivluscle contraction affects stress patterns in bone by producing compressive stress that partially

or totall).' neutralizcs thc tcnsilc stress acting on the bone.Bone is stiffer, sustains higher loads before fail- ing, and stores more energy whcn loaded at higher physiological strain rates.

~i Living bone fatigues when the frequcncy.' of loading precludes the remodeling necessary to pre- vent failure.

The mechanical behavior of a bone is influ- enced by its geometry (length, cross-sectional area, and distribution of bone tissue around the neutral

Bone remodels in response to the mechanical demands placed on it; it is laid down \vhere needed and resorbed where not needed.

vVith aging comes a marked reduction in the amount of cancellous bone and a decrease in the thickness of cortical bone. These changes diminish bone strength and stiffness.

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C VASCULAR FACTO,RS J

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Protection, suppOrt. kinCffiJtic links

MECHANICAL FACTORS

Function

SlruCturc

;,N~~coliag~n'Proteln~; Growth factor cell attachment proteins PG',

FLOW CHART 2·2

~This flow chart is dl?~igncd for classroom or group discussion. Flow chart is not meant to be exhaustive.

~~ Organic Component \1

Collagen Type Proteins I ) 1 Bone composition. structure, and functions.· *(PG's,* proteoglycans)

Inorganic Component

Cry,Olh of HYDROXYAPATITE '

Calcium, Phosphate

~~. ;~:: to6s: 'm() I}>Zn'"o-;Vi '"Cm'"~o'"-<'"C()-;C'"m'"o-; Im~'"()Cro'""mr'm~r'"-<~ms:

• a. .. .".s V '" E m x ~ ~ .0 "' .~ ·u V v E '" c0 '" '" m E mm**"a** " > 0£ 0 "0 .0 0 B c m0 E <; c {; .- em u: 0 ~ **":;:** C0 'i3 ~ 0 m ~ § ~ 0 2 '" 5 0 .:" u E 0 .;;;u **"5** w C x .e .,u "".-m 0c'"~ •'""'~;;;0~

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FLOW CHART 2-4 Intrinsic factors associated with bone damage. Clinical examples.\*

"This flow chart is designed for d<lssroom or group discussion. Flow chart is not meant to be cxhamtivc.

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.Ostco~h'~n'd~i;i'~ disseC30SKcil:, , .Os,conccrosi~~~ of the condrlos'\0" femoral , "';; *j:,*

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PATHOLOGICAL OR FRAGILITY FRACTURE

Tumoral DIsturbance Gcnedc DIsturbance

\' .',,-. ,,,,:,)::,::~,'.i:; ·Gauchcrs disease

·Cushing Syndrome

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Biomechanics of **Articular Cartilage**

*Van* C. *Mow, Clark T Hung*

Introduction

Composition and Structure of Articular Cartilage

Collagen Proteoglycan WatE'r Structural and PhysicallniE'raction Among Cartilage Components

Biomechanical Behavior of Articular Cartilage Nature of Articular Cartilage Viscoelasticity Confined Compression Explant Loading Configuration Biphasic Creep Response of Articular Cartilage in Compression Biphasic Stress-Relaxation ResponsE' of Articular Cartilage in

Compression Permeability of Articular Cartilage Behavior of Articular Cartilage Under Uniaxial Tension Behavior of Articular Cartilage in Pure Shear Swelling Behavior of Articular Cartilage

Lubrication of Articular Cartilage

Fluid-Film Lubrication Boundary Lubrication Mixed Lubrication Role of Interstitial Fluid Pressurization in Joint Lubrication

Wear of Articular Cartilage

Hypotheses on Biomechanics of Cartilage Degeneration

Role of Biomechanical Factors Implications on Chondrocyle Function

Summary

Acknowledgments

References

Flow Charts

J

*Composition Introduction*

*and Structure* of *Articular Cartilage* Three types of joints exist in the human body: fi- b'ro~ls, these,;-,ollle cartilagin,olls. syn.ovial. or and diarthrodial, synovial. jo.int, Onl.y allows one Chondrocytcs, the sparsely distributed cells in artic- lllarc~\I:tiL\ge,account for less than 10% of the tis- sue's volume (Stoekwell. 1979). Schematically. Ihe zonal arrangement of chondrocytcs is shown in Fig- ure 3-1. Despite their sparse distribution, chondro- cytcs manufacture, secrete, on!anil',---c, and rnainlain tile on!anicco'n;-I;~I~l-~f the~extracellular matrix (ECM)(Fos'lIlg & Hardingham, 1996; ·M~lir, 1983). The org~nic ma~rix,is compo~~5:L~~[.:~,,(I(:nscp9.J~\'ork of miii6i' fin:"-§?·I-(..1¥8~,,,(~'bril~ amounts .,;1' types (n10stly V, VI, t'ype-Jl~c-olh;~gen.,with IX, and XI) that "~reo cans(pcs) enrneshcd..Jn. (l3ate;;'an a cOnCC.tl.traled et aI., ~olu.~i().n. of proteogly- I996;Eyre, 1980; Mloir, -~1983)\_ 1n nOITn,)1 arLiculaL\_cartilage the collagen cOlllent ranges from 15 to *22%* by wet weight and the PG con-tentfronl4to maining 60 to *85%* is 7% walCI: bv .' inorganic wet wei(Tht· :::::> salts. ' the re- and small amounts of other matrix proteins, glycopro- teins. and lipids (Mow & Ratcliffe, 1997). Collagen flbrils and PGs, each being capablc of forming structural nctworks of significant strength (Broom & Silyn-Roberts, 1990; Kempson et aI., 1976; Schmidt el aI., 1990; Zhu et aI., 1991, 1993), arc Ihe structural components supporting the internal me- chanical stresses that result rrom loads being ap- plied to the articular canilage, Moreover, these structural components, together with water, deter- mine the biomcchanical bchavior of this tissue (Ateshian et aI., 1997; Maroudas, 1979; Mow et aI., ~rticular 1980, 1984; Mow & Aleshian, 1997).

A notable exception to the definition of hyaline articu- lar G'lrtilage is the temporomandibular joint, a synovial

Collagen is the mosl abundant protein in the body

joint in which fibrocartilage is found covering the bone

(Bateman et aI., 1996; Eyre, 1980). In articular carti-

ends. Fibrocartilage and a third type of cartilage, elastic

lage. collagen has a high level of structural organiza- cartilage. are closely related to hyaline cartilage embry-

tion that providcs a fibrous ultrastructure (Clm-k, ologically and histologically but .'lre vastly different in

1985; Clarke, 1971; Mow & Ratcliffe, 1997). The ba-

mechanical and biochemical properties. Fibrocartilage

sic biological unit of collagen is tropocollagen. a

represents a transitional cartilage found at the margins

Sln.lcture composed of three procollagen polypeptide of some joint cavities, in the joint capsules, and at the

chains (alpha chains) coiled into left-handed helixes

insertions of ligaments and tendons into bone.

(Fig. 3-2;\) thaI are furthcr coiled abollt each other Fibrocartilage also forms the menisci interposed be-

inlO a right-handed triple helix (Fig. 3-28). These

tween the articular cartilage of some joints and com-

rod-like tropocollagen molecules. 1.4 nanOJlletcl-s poses the outer covering of the interverlebral discs. the

(nm) in diameter and 300 nm long (Fig. 3-2, C & *0),*

anulus fibrosus. Elastic cartilage is found in the external

polymerize into larger collagen fibrils (Bateman

ear, in the cartilage of the eustachian tube, in the

ct al.. 1996; Eyre, 1980), In articular cartilage, these epiglouis. and in certain pans of the larynx.

fibrils have an average' diamcter of 25 to 40 nm (Fig.3-2E, Box 3-2); howevec this is highly variable. of' a large degree of motion. In Y~\_L1ng normal joints, the ,-~llticulalingJ)onecn~ls of diartl~~'odiaIj9\_inls,!.re cov- }--ered by a-thin (1-6 mill), den~c, tram~JJlcc\_nt,~:-,'hile ,~~c6nnective-i.i.ssllc iCBox-3:!). Articul'IU~f\rtilage-;"s called hyaline arliCltlar. cartilage a highly s-p·ecia·lized .tissue precisely suited for withstanding the highly i~adcd joint environment without failure during an a\lerage individual's lifetime. PhXti.t~~l.Qgically, how- ever, ,il is v51~~~I'.lIly an isolated li,ssuc, devQ.id of blood \;essds, 1).~-mphalic channels, and nelyrol.ogical inncr- ~alion~'-Flll'lllermoJ'c,its .c~l.llliar density is less than .·tK"t of any other tissue (Stoek\~c·il. 1979).

[n (liartll-i'o(Ti~;fioints, articular cartilage has two - p-rimar'yfl~nctions': (1) to distl-ibute jo.int loads over ,a wiele al-ea, thus decreasing lhe strc:)ses sllst~!.!ned by the contacting join.t..~.sl~rraccs (Atcshian ct aI., 1995; Helminen et aI., 1987) and (2) to allow rel"tive movementoft!1cu:m,posing joinl surfaces \~:·,i.t"I,l""I~lin­ imal ft:'iction and wear (iVlow & Alcshian, 1997). In thi's"ch~l'I)iel','wcw'Hi describe how the biomcchani- cal pl'openics of articular cartilage, as determined by mal its performance composition or and these structure, functions.

allow for the opti-

I,Cartilage 61

~.~- ..,

COLLAGEN

Articular ...... ,,,,'•.,,•.,0' surface

STZ (10.20%)

., ,,~) •..,.> \_,~ Middle zone (40-60<::0) ," .~ ,., '. '~,,~-' ,,-r:~ '~

,~.-r ,'-' ~{~:f2jJ.~ Deep zone (30%) i~;~f Calcified zone

";·'5c..C\_~,.,,1::;\,,\_, Subchondral bone

B

" Tide mark Chondrocyte Photomicrograph (A) and schematic representation (8) of the chondrocyte arrangement throughout the depth of noncalcified articular cartilage. In the superficial tangential zone, chondrocytes are oblong with their long axes aligned parallel to the articular surface. In the middle zone, the chondrocytes are "round" and randomly distributed. Chondrocytes in the deep zone are arranged in a columnar fashion oriented perpendicular to the tidemark,

**•** the demarcation between the calcified and noncalcified tissue.Scanning electron microscopic studies, for instance, have described fibers with diameters ranging up to 200 nm (Clarke, 1971). Covalent cross-links form be- tween these tropocollagen molecules, adding to the flbrils high tensile strength (Bateman et aI., 1996).

The collagen in articular cartilage is inhomoge- neously distributed, giving the tissue a layered char- acter (Lane & Weiss, 1975; Mow & Ratc1iITc, 1997). Numerous investigations Llsing light. transn1ission electron, and scanning electron microscopy have identifled three separate structural zones. For ex- ample, Mow et al. (1974) proposed a zonal arrange- ment for the collagen network sho\vn schematically' in Figure *3-3A.* **In** the superficial tangential zone, which represents 10 to 20()0 of the total thickness, are sheets of fine, densely packed fibers randomly \voven in planes parallel to the articular surface (Clarke, 1971; Redler & Zimny, 1970; Weiss et aI., 1968). In the middle zone (40 to 60% of the total thickness), there are greater distances between the

Differences in Coliagen Types

randomly oriented and homogeneousl::... dispersed fibers. Below this, in the deep zone (approximatel.v

Differences in tropocollagen alpha chains in various *30%* of the total thickness), the fibers come together,

body tissues give rise to specific molecular species, or forming larger, radially oriented fiber bundles

types of collagen. The collagen type in hyaline cartilage, (Redler et aI., 1975). These bundles then cross the

type II collagen, differs from type I collagen found in tidemark, the interface between articular cartilage

bone, ligament, and tendon. Type II collagen forms a and the calcified cartilage beneath it, to enter the

thinner fibril than that of type I, permitting maximum calcified cartilage, thus forming an interlocking

dispersion of collagen throughout the cartilage tissue. "root" sj'stem anchoring the cartilage to the uncler-

lying bone (Bullough & Jagannath, 1983; Redler et aI., 1975). This anisotropic fiber orientation is mir- rored by the inhomogeneous zonal variations in the collagen content, which is highest at the surface and then remains relatively constant throughout the deeper zones (Lipshitz et aI., 1975). This composi- tionalla.vering appears to provide an important bio- mechanical uniformly across function the by' loaded distributing regions the or the stress joint more tis- sue (Selton et aI., 1995).

Cartilage is composed primarily of type II colla- gen. [n addition, an array of difFerent collagen (t)'Pes V, VI. IX, Xl) can be found in quantitatively minor amounts within articular cartilage. Tvpe II collagen is present primarily in articular cm·tilage, the nasal septum, and sternal cartilage, as well as in

(r chain

A

Bo

1.4 I l

nm Triple helix

Tropocollagen molecule

Collagen fibril with quarter-stagger array of molecules

Fibril with repeated banding pattern seen under eleclron microscope

Molecular features of collagen structure from the alpha chain ((I) to the fibril. The flexible amino acid sequence in the alpha chain (A) allows these chains to wind tightly into a right- handed triple helix configuration (8), thus forming the tropocollagen molecule (C). This tight triple helical arrangement of the chains contributes to the high tensile strength of the collagen fibril. The parallel alignment of the individual tropocollagen molecules, in which each molecule overlaps the other by about one quarter of its length (0), results in a repeating banded pattern of the collagen fibril seen by eledron microscopy (x20,OOO) (E). *Reprinted wirh permission from Donohue, It'l1., Buss. D., Oegema. IR., et al.* (19831- *The effects of indireCl blunt trauma* Oil *adult canine arricuhu cartilage.* J Bone Joint Surg, GSA. 948.

**•**

the inner regions of the intervertebral disc and meniscus. For reference, type I is the most abun- dant collagen in the human body and can be found in bone and soft tissues such as intervertebral discs (rnainl.y in the annulus fibrosis), skin. meniscus, ten- dons, and ligaments. The most important mechani- cal properties of collagen fibers nrc their tensile stiffness and their strength (Fig. 3-4;\). Although a Single collagen fibril has'-not be~n tested in len;ion. the tensile slI'ength of collagen can be inferred from tests on structures with high collagen contenl. Ten- dons, for example, are about 80% collagen (dry weight) and have a tensile stiffness of 10.1 !\'1Pa and a tensile strength of SO MPa (Akizuki et aI., 1986; Kempson, 1976, 1979; Wooet aI., 1987, 1997).Stec1, by comparison, has a tensile stillness of approxi- mately 220 x 10" MPa. Although strong in tension. collagen fibrils offer little resis(ance lo ~olllpression

because their large slendcnlcss ratio, the ratio of length to thickness, makes it easy for them to buckle under cotl1prcssive loads (Fig. 3-48).

Like bone, articular canilage is anisotropic; its material properties differ with the direction of loacl~ ing (Akizuki et aI., 1986; Kempson, 1979; Mow & RalclifTc, 1997; Roth & Mow, 1980; Woo el al.. 1987). Oft is thought that this anisotropy is related to the varying collagen fiber arrangements within the planes parallel to the articular surface. It is also thought, however, that variations in collagen f-iber cross~link density, as well as variations in collagen- PG interactions. also contribute to articular carti- lage tensile anisotrop~", In tension, this anisotropy is llsually described with respect to the direction of the articular surface split lines. These split lines arc elongated fissures Jj·roduced by piercing the ar- ticular surface with a small round awl (Fig. 3-5;

O'TMW"PAMH;.

B STL Middle zone Deep zone

'1:;<,.'*.J*'kf"'4,

A. Schematic representation, *(Repriflleej \·virh permission from Mow* v.c. *el al. j1974j. Some surface dJc1riJcteristics of arriculaf cartilages. A scanning electron microscopy sludy and* a *theoretical mode! for the dynamic interaction of synovial fluid and articular car· tilage.* J Bionll?{hanics, 7, 449), B. Photomicrographs (x3000; provided through the courtesy of Dr. T. Takei, Nagano. Japan) of the ultrastructural arrangement of the collagen network throughout the depth of articular cartilage. In the superficial tangential zone (STZ), collagen fibrils are tightly woven into sheets arranged parallel to the articular surface. In the middle

zone, randomly arrayed fibrils are less densely packed to ac- commodate the high concentration of proteoglycans and wa- ter, The collagen fibrils of the deep zone form larger radially oriented fiber bundles that cross the tidemark, enter the calci· fied zone, and anchor the tissue to the underlying bone. Note the correspondence between this collagen fiber architecture and the spatial arrangement of the chondrocytes shown in Figure 3-1. In the above photomicrographs (B), the STZ is shown under compressive loading while the middle and deep zones are unloaded.

<)0

1m)) Collagen )))))) 1m)) Fibril

)l»ll IllJli lJIlil I(> AHigh tensile sliflness and strength B Uttle resistance to compression

Human femoral condyles

Illustration of the mechanical properties of collagen fibrils: (A) stiff and strong in tension, but (B) weak and buckling easily with compression. *Ad,lpted from Myers, E.R., Lai, WM., & MOV'I, VC* (1984). *A COfltinuum theory and* all *experiment for the ion-induced swelling be/Mvior carlilage.* j Biomech Eng, 106(21. 15/-/58.

Diagrammatic representation of a split line pattern on the surface of human femoral condyles. *Reprinted wirh permis- sion from Hu/rkrantz,* W (1898). *Ueber die Spafrrichwflgefl der Gelenkknorpel.* Verhandlungen der Analomischen Gesellschaft, 12.248.

ARTIC0lJ'.~F~~rlf'~f1k'l!~.

Hultkrantz, 1898). The origin of the pattcrn is re- lated to the directional variation 01" the tensile stilT- ness and strength characteristics of articular cani- lage described above. To dale. ho\Vevel~ the exact rc;sons as to why articular canilage exhibits slIch pronounced nor is the functional anisolropies significance in tension or is not this known, tensile anisotrop)'.

binding region in c.:anilagc. iVlore recently, the other two globular regions have been extensively studied (Fosang & Hanlingham. 1996), but their functional significance has notycl been elucidated. Figure 3-68 is lhe accepted molecular confonnation of a PG ag- gregate; Rosenberg Gt al. (1975) were the first to obtain an electron micrograph 01" this molecule (Fig. *3-6C).*

PROTEOGLYCAN

l\llany types of PGs arc round in cartilage. Funda- mentally, it is a large protein-polysaccharide mole- cule composed of a protein core to which one or more glycosaminoglycans (GAGs) are attached (Fosang & Hardingham, 1996; Muir, 1983; Ratcliffe & Mow, 1996). Even the smallest of these molecules, high'Can and deem-in, are quite large (approxi- m;t~ly I X 10'\ 111\\'). but they comprise less than 10~~ of all PGs present in the tissue. Aggrccans are much larger (1-4 X 10" mw), and they have the rc- markable capability to attach to a hyaluronan mol· ccule (HA: 5 X 10' I11W) via a specific 1·-1A-binding rc- gion (HABR). This binding is stabilized by a link protein (LP) (40-48 X 10" mw). Stabilization is cru- cial 10 the function of normal cartilage; without it, the components of the PG molecule would rapidly escape from the tissue (Hardingham & IVluir, 1974; H<lscall, 1977; Muil; 1983).

Two types of GAGs comprise aggrecan: chon- droitin sulfatc (CS) and kcmtan sulfate (KS). Each CS chain contains 25 to 30 disaccharide units, while the shorter KS chain contains 13 disaccharide units (MuiJ~ 1983). Aggrecans (previously rererred to as subunits in the American literature or as monomers in the UK and European literature) consist ofan ap- proximately 200.nanol11ctcr..long protein core to which approximately 150 GAG chains, and both 0- linked and N-linked oligosaccharides, are covalently attached (Fosang & Hardingham, 1996; Muir, 1983). Furthermore, the distribution of GAGs along the protein core is heterogeneous; there is a region rich in KS and O-linked oligosaccharides and a rcgion rich in CS (Fig. *3-6M.* Figure 3-611 dcpicts the fa- mous "boule-brush" model for an aggrecan (j\lluir, 1983j. Also shown in Figurc 3-6/\ is the hctcrogcne- ity of the protein core that contains three globular regions: GIl the HABR located at the N-terminus that contains a small amount of KS (Poole, 1986) and a the few HABR- N-Iinked and oligosaccharidcs, the KS-rich regio·n G., located (Hardingham between et aI., 1987), and G" the corc protcin C-tenninus. A I: I stoichiomelly C:dslS between the LP and the *GI*

In native cartilage, most aggrccans arc associated withHA to form the large PG aggregates (Fig. *3-6C).* These aggregates may have up to several hundred aggrecans noncovalenliy uuached to a central HA core via their HABR, and each sile is stabilized by an LP. The filamentous HA core molecule is a non- sulfated disaccharide chain that may be as long as 4 fJ.m ill Icngth. PG biochcmists have dubbed thc HA an "honorary" PG, as it is so intimately involved in the structure of the PG aggregate in articular carti- lage. The stability afforded by the PG aggregates has a rl"wjor functional significance. It is accepted now thut PG aggregation promotes immobilization of the PGs within the nne collagen meshwork, adding structural stability and rigidity to thc ECM (Mow ct aI., 1989b; Muir, 1983; Ratcliffc et aI., 1986). Fur- thermore, two additional forms of c!crmatan sulfate PG have been idcntif"led in the ECM of articular car- tilage (Rosenberg et aI., 1985). In tendon, c1ermatan sulfate PGs have been shown to bind noncovalently to the surfaces of collagen fibrils (Scott & Orford, 1981); however, the role of dermatan sulfatc in ar- ticular cartilage is unknown, biologically and func~ tionally.

Although aggrecans generally have the basic structure as described abovc, they arc not struc- turally identical (Fosang & Hardingham, 1996). Ag- grecans vary in length. molecular weight, and com- position in a variety of ways; in other words, they arc polydisperse. Studies have demonstrated two distinct populations of aggrecans (Buckwalter et aI., 1985; Heincgard ct aI., 1985). The first population is present throughout life and is rich in CS; the second contains PGs rich in KS and is present only in adult cartilage. As articular cartilage matures, other agc- related changes in PG composition and structure occur. vVith canilage maluration, the watcr content (Armstrong & Mow, 1982; Bollet & Nancc, 1965; Linn & Sokoloff, 1965; Maroudas, 1979; Venn, 1978) and the carbohydrate/protein ratio progres- sively decrease (Garg & Swann, 1981; Roughley & White, 1980). This dccrcase is mirrored by a de- crease in the CS content. Conversely, KS. which is present only in small ,imounts at birth, increases throughollt development and aging. Thus, the

C-terminal globular domain (G3)

AB

Proteoglycan (PG) Macromolecule

1200 nm

1DI1I'---- \_

Hyaluronan (HA)

iIfII!!!...""". ,IIIIIIIIII!!I,II,'IIIIlIl!!.,!,!!!,,.p.~",,~~",•."!!"!'!.~""~"::-'~••••~·,.cT·","'!'!'.,"'.,=-,~ .. "'-.----:'....,....-----~.

A. Schematic depiction of aggrecan, which is composed of keratan sulfate and chondroitin sulfate chains bound cova- lently to a protein core molecule. The proteoglycan protein core has three globular regions as well as keratan sulfate- rich and chondroitin sulfate-rich regions. B. Schematic rep- resentation of a proteoglycan macromolecule. In the ma- trix, aggrecan noncovalently binds to HA to form a macromolecule with a molecular weight of approximately 200:<. 106. Link protein stabilizes this interaction between the binding region of the aggrecan and the HA core mole- cule. C. Dark field electron micrograph of a proteoglycan aggregate from bovine humeral articular cartilage *(>:120,000). Horizontal fine* at lower right represents O,SlAm. *Reprinted with permission from Rosenberg,* C, *Hellmann, W, & Klein5chmidr. AX (975). Electron microscopic studies* 0; *proteD- gfycan aggregates (rom bovine articular cartifage.* J Bioi Chem, 250. 1877.

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Glycosaminoglycan (GAG) chains (KS. CS)

Chondroitin Sulfate (CS) chains

*f '*

*CS/KS* ratio, which is approximately 10: 1at birth, is only approxirnatel:v 2: I in adult cartilage (Roughlcy & "Vhile, 1980: Sweet et al" 1979; Thonar et al" 1986). Furthermore, sulfation of the CS molecules, \vhich can occlir at either the 6 or the 4 position, also undergoes age-related changes. In utero, chon- droitin-6-sulfate and chondroitin-4-sulfate are pres- ent in equal molar amounts; however, by maturity, the chondroitin-6-sulfate:chonclroitin-4-sulfatc ra- tio has increaseel to approximately 25: I (Roughley et a1., 1981). Other studies have also documented an age-related decrease in the hydrodynamic size of tf~e aggrecan. Many' of these early changes seen in articular cartilage may reflect cartilage maturation, possibly as a result of increased functional demand \vith increased \vcight-bcaring. Ho\vever, the func- tional significance of these changes, as well as those occurring later in life, is as .\'ct undetermined.

WATER

\Valel~ the most abundant component of articular cartilage, is n10st concentrated ncar the anicular surface *(\_80°;(j)* and decreases in a near-linear fash- ion with increasing depth to a concentration of ap- proximately 65% in the deep zone (Lipshitz et al., 1976: Maroudas, 1979). This Iluid contains many free mobile cations (e.g., Nu', K', and Cal.) that greatly! influence the mechanical and physicochem- ical behaviors of cartilage (Gu et aI., 1998; Lai et aI., 1991; Linn & Sokoloff, 1965; Maroudas, 1979). The fluid component of articular cartilage is also essen- tial to the health of this avascular tissue because it permits gas, nutrient, and waste product movement back and forth between chondrocytes and the sur- rounding nutrient-rich synovial fluid (Bollet & Nance, 1965; Linn & Sokoloff, 1965; Mankin & Thrashel; 1975; Maroudas, 1975, 1979).

A small percentage of the \vater in cartilage re- sides intracellularly', and approximately *30%* is strongly associated with the collagen fibrils (Maroudas et aI., 1991; Torzilli et aI., 1982). The in- teraction between collagen, PG, and water, via Don- nan osmotic pressure, is believcd to have an im- portant function in regulating the structural organization of the ECM and its s\velling properties (Donnan, 1924; Maroudas, 1968, 1975). Most of the \vater thus occupies the interfIbrillar space of the ECM and is free to move whcn a load or pressure gradient or other electrochemical motive forces are applied to the tissue (Gu et aI., 1998; Maroudas, 1979). When loaded by a compressive force, ap-

proximately *70(;r;* of the water may be moved, This interstitial fluid movement is important in control- ling cartilage mechanical behavior and joint lubri- cation (Ateshian et aI., 1997, 1998; Hlavacek, 1995; Hou et aI., 1992; Mow et aI., 1980; Mow & Ateshian, 1997).

STRUCTURAL AND PHYSICAL INTERACTION AMONG CARTILAGE COMPONENTS

The chemical structure and physical interactions of the PG aggregates influence the properties of the ECM (Ratcliffe & Mow, 1996). The closely spaced (5-15 angstroms) sulfate and carboxyl charge groups on the CS and KS chains dissociate in solu- tion at physiological pH (Fig. 3-7), leaving a high concentration of fixed negative charges that create strong intramolecular and intermolecular charge- charge repulsive forces; the colligative sum of these forces (when the tissue is immerscd in a physiological saline solution) is equivalent to the Donnan osmotic pressure (Buschmann & Grodzin- sky, 1995; Donnan, 1924; Gu et aI., 1998; Lai et aI., 1991). Structurally, these charge-charge repulsive forces tend to extend and stiffen the PG macro- molecules into the interfibrillar space formed by the surrounding collagen network, To apprcciate the magnitude of this force, according to Stephen Hawkings (1988), this electrical repulsion is one million, million, million, million, million, million, million times (42 zeros) greater than gravitational forces.

In nature, a charged body cannot persist long without discharging or attracting counter-ions to maintain electroneutrality, Thus, the charged sulfate and carboxyl groups flxed along the PGs in articular cartilage must attract various counter-ions and co- ions (mainly Na', Cal., and CI") into the tissue to maintain electroneutrality. The total concentration of these counter-ions and co-ions is given by the well-known Donnan equilibrium ion distribution law (Donnan, 1924). Inside the tissue, the mobile counter-ions and co-ions form a cloud surrounding the fixed sulfate and carboxyl charges, thus shielding these charges from each othet: This charge shielding acts to diminish the very large electrical repulsive forces that otherwisc would exist. The net result is a swelling pressure given by the Donnan osmotic pres- sure law (Buschmann & Grodzinsky, 1995; Donnan, 1924; Gu et aI., 1998; Lai et aI., 1991; Schubert & Hamerman, 1968). The.Donnan osmotic pressure theory' has been extensivel:v used to calculate the

Aggregate Domain

Repulsive forces due 10 fixed charge density distribution A

Smaller domain

Increased Increased charge charge-charge density repulsive forces BA, Schematic representation of a proteoglycan aggregate solution domain *(left)* and the repelling forces associated with the fixed negative charge groups on the GAGs of ag- grecan *(right).* These repulsive forces cause the aggregate to assume a stiHly extended conformation, occupying a large solution domain, B. Applied compressive stress de- creases the aggregate solution domain *(left),* which in turn increases the charge density and thus the intermolecular charge repulsive forces *(right).*

swelling pressures of anicular cartilage and the intervertebral disc (Maroudas, 1979; Urban & McMullin, 1985), By Starling's law, this swclling pressure is, in llll'n. resisted and balanced by tension developed in the collagen network, confining the PCs to only 20% of their fTee Solulion domain (Maroudas, 1976; Mow & Ratcliffe. 1997; Sellow eL aI., 1995). Consequently. this swelling pressure subjects the collagen network to a "pre-stress" of sig·

nificant rnagnitudL: c\'en in the absclH.':c or cxternal loads (Sellon cl aI., 1995, 1998).

Canilage PGs arc inhomogeneollsl.\' dislribuled th,'oughoul the malrix, with their concenlralion gell- erally being highest in the middle zone and lowest in the superficial and deep zOlles (Lipshitz Cl aI., 1976; ,'Jlaroudas, 1968, 1979; Vcnn, 1978). Thc biomechan- ical consequenceor this inhomogeneous swelling be- havior of cartilage (caused b.v the varying PG content throughout the depth of the tissue) has recently been quantitalively assessed (Sellon el al., 1998), Also, re- sults from n:cent finite clement calculations based on models incorporating an inhomogeneous PG dis- lribution show lhat it has a proround effect on the interstitial counler-ion dislribulion throughoul the depth of the lissuc (Sun et aI., 1998).

\·Vhen a compressive stress is applied lo the car- lilage surface. there is an inSlantaneous deforma- lion caused primal"i)~' by a change in the PG mol- ecular domain, Figure 3-7*B.* This cxtc.'rnal stress causes the inlernal prcsstll"e in the matrix to ex- ceed the swelling pressure and thus liquid will be- gin to flow out of" tht: tissue. As the fluid r!()\VS out, the PG concentration increases, which in turn in- creases the Donnan osmotic swelling pressure or the charge-charge repulsive force and bulk com- pressive stress unlil they are in equilibrium with the external stress. **In** this manner, the physico- chcmical properties of the PC gcl lrapped wilhin the collagen network enable il to resisl compres- sion. This mechanism complements the role pla~:ed by collagen that, as previollsly described, is strong in lC'nsion but wcnk in compression, The abilit.'· of PCs to resist compression thus arises fl'om two sources: (I) the Donnan osmotic swelling pressure associated with the tightly packed fixed anionic groups on the GAGS and (2) the bulk eomprcssivc stillness of the collagen- PG solid matrix, Experimentally, the Donnan os- motic pressure ranges from 0.05 to 0.35 MPa (Maroudas, 1979), while the elastic modulus of the collagen-PG solid matrix ranges from 0.5 to 1.5 MPa (Armstrong & Mow. 1982; Alhanasiou et al.. 1991; Mow & Ratcliffe, 1997).

Il is now apparenl that collagen and PGs also interact and thal these interactions m'e of great functional imponance. A small portion or the PGs have been shown to be closely associated with col- lagen and may serve as a bonding agent belween lhe collagen fibrils. spanning distnnccs thLit ::H'C too great for collagen cJ."oss·links to develop (Bateman ~t al.. 1996; Mow & Ratcliffe, 1997; Muir, 1983).

PGs arc also thought to play an important role in maintaining the ordered structure and mechani- cal properties of the collagen fibrils (Muir, 1983; Scott & Orford, 1981). Recent investigations show that in concentrated solutions, PGs interact with each other to form networks of significant strength (Mow et aI., 1989b; Zhu et aI., 1991, .i996)..Morcover, the density and strength of the interaction sites forming the network \vere shown to depend on the presence of LP between aggre~ cans and aggregates, as well as collagen. Evidence sl.w:gests that there are fewer aggregates, and m(;~e biglycans and decorins than aggrecans, in the superficial zone of articular cartilage. Thus, there mllst be a difference in the interaction be- l\VCCn these PGs and the collagen fibrils from the superficial zone than from those of the deeper zones (Poole et aI., 1986). Indeed, the inle"action bet\veen PG and collagen not only' plays a direct role in the organization of the ECr.,ll but also con- tributes directly' to the mechanical properties of the tissue (Kempson el aI., 1976; Schmidt el aL. 1990; Zhu el aI., 1993).

The specific characteristics of the physical. chemical. and mechanical interactions between coHagen and PG have not yet been fully' deter· mi.ned. Nevertheless, as discussed above, we know that these structural macromolecules interact to form a porous·permeable, fiber-reinforced com- posite matrix possessing all the essential mechani- cal characteristics of a solid that is swollen with \vat.er and ions and that is able to resist the high stresses and strains of joint articulation (Andriacchi etal.. 1997; I-lodge el aI., 1986; Mow & Ateshian, 1997; Paul, 1976). It has been demonstraled lhat these collagen-PG interactions involve an aggre- call, an I-IA filan1ent, type II collagen, other minor collagen types, an unknown bonding agent, and possibly smaller cartilage components such as col- lagen type IX, recently identified glycoproteins, and/or polymeric HA (Poole ct aI., 1986). A schematic diagram depicting the structural arrangement \vithin a small volume of articular cartilage is shown in Figure 3-8.

\Vhen articular cartilage is subjected to external loads, the collagen-PG solid matrix and interstitial lIuid function together in a unique way to protect against high levels of stress and strain develop- ing in the ECM. Furthermore, changes to the bio- chemical composition and structur,,;l organization of lhc ECM, such as during osteoarthritis (OA), are paralleled b:y changes to t';e biomechanical proper-

Hyalur n

-'I,-\-----Interstltlal flUid

Collagen fibril

Schematic representation of the molecular organization of cartilage, The structural components of cartilage, collagen, and proteoglycans, interact to form a porous composite fiber-reinforced organic solid matrix that is swollen with water. Aggrecans bind covalently to HA to form large pro- teoglycan macromolecules.

ties of cartilage. In the following section, the be- havior of articular cartilage under loading and the mechanisms of cartilage fluid flow will be discussed in detail.

*Biol71echanical Behavior* o{ *Articular Cartilage*

The biomechanical behavior of articular cartilage can best be understood when the tissue is viewed as a multiphasic medium. In the present context, artic- ular cartilage will be treated as biphasic material consisting of two intrinsically incompressible, im- miscible, and distinct phases (Bachrach et aI., 1998; Mow et aI., 1980): an inlerstitial fluid phase and a porous-permeable solid phase (i.e., the ECM). For explicit analysis 01' the contribution 01' the PG charges and ions, one would have to consider three distinct phases: a fluid phase, an ion phase, and a charged solid phase (Gu el al.. 1998; Lai et aI., 1991). For understanding how the water contributes to its

mechanical propcnies, in the prc.:SCnl context. arti<,:. lIlar cartilage may be considered as a Ouid-filled porous-permeable (uncharged) biphasic medium, with each constituent playing a role in the functional behavior of cartilage.

During joint arlicuhllion, forces at the joint sur- face may vary" from alnl0S1 zero to more than ten times body weight (Alldriacchi et ai., 1997; Paul, 1976). The contact arcas also vary in a cOll1plcx manner and typically they arc only or the order of several square centimeters (Ahmed & Burke. 1983; Ateshian et al.. 1994). It is estimated that the peak contact stress may reach 20 MPa in the hip while rising from a chair and to IVIPa during stair climb- ing (Hodge et aI., 1986; Newberrv et aI., 1997). Thus, articular cartilage, under physiologicallonding con- ditions, is a highly stressed malcrial. To understand how this tissue responds LInder these high physio- logical loading conditions. its intl"insic mechanical properties in compression, tension, and shear must be determined. From these properties, one can un· dcrstand the load-can~ying mechanisms within the ECM. Accordingly, the following subsections \\"ill characterize tht: tisslIC' behavior under these loading Il'lOclalities.

NATURE OF ARTICULAR CARTILAGE VISCOELASTICITY

If a material is subjected to the action of a constant (time-independent) load or a constant deforrnation and its response varies with time, then the mechan- ical behavior of the material is said to be viscoelas- tic. **(n** general, the response of slich a material can be theoretically modeled as a cornbination of the re- sponse of a viscolls Ouid (dashpot) and an elastic solid (spring), hence viscoelastic.

The two fundamental responses of a viscoelaslic material arc creep and stress relaxation, Creep oc- curs when a viscoelastic solid is subjected to the ac- tion of a constant load, Typically. a viscoelastic solid responds with a rapid initial deforrnation followed by a slow (time-dependent), progressivel)' increas- ing deformation known as creep until an equilib- rium state is reached. Stress relaxation occurs when a viscoelastic solid is subjected to the action of a constant deformation, l~ypically. a dscoelastic solid responds with a rapid, high initial stress followed by a slow (time~dependent). progressively decreasing stress required to maintain the deformation; this phenomenon is known as stress relaxation.

Creep and strc~s rL,lnxalion phenomcna ll1a~' be caused b~' differenl Illcchani:sms, For single-phase solid pol~'lTh:ric materials, these phenomena are the result of iJHcrnal friction caused b.\' the mOlion or the long polymL'ric chains sliding over ('ach olher \\'ithin the stressed lll~lIcrial (Fling, 1981), The viscocl"stic bdmvior of tendons and ligamt::IHs is primarily' eauscd bv this mechanism (Woo et aI., 1987, 1997). For bone, the long-term viscoelastic lK'hador is thought to ilL' caused h.\' a rdati\'c slip uf lamellae within the osteons along with the Ilow of the inlL'rsti~ tial Iluid (Lakcs & Saha, 1979). For articul,,,' carti- hlgC. the compressive \'iscoelastic bdHl\'ior is prilllar~ i1~' caused h~' the Jlo\\' of tilL' interstitial lillie! and lhe frictional drag associated with this lIow (Ateshian cl aI., 1997; Mow (,t aI., 1980, 1984). In shear, as in single-phase viscoelastic pol.vmers. it is primarily c<.lusL·d b~' tilt:' motion of 10l1g pol~!J1lL'r chains such as collagen and PGs (Zhu d aI., 1993, 1996). Thc com- ponent or anicu[;.\r cartilage dscoclasticit~,caused by interstitial lluid !low is known as the hiphasic vis- cot.,lastic behavior (i'VlO\\' et a!., 1980), ~lI1d the COI11- ponent of dscoebslicil~'caused b~' macromolecular motion is known as til(' l1ow-indepl'ndelll (Ha,\'cs '\* Bc)dinc, 1(78) or the intrinsic visc()elastic behavioror Ihe collagen-PG solid matrix.

Although the deformational bdwyior has been described in terms of n lillem' elastic solid (Hirsch, 1944) or viscoclastic solid (I-laves & Mockros, 1971), lhes~ modcls fail to recognii'.e the role or water in the \'iscoelastic behavior or and thc significant con- tribution lilal lIuid pressurizalion pla.vs in joint load support and canilage lubrication (Ateshian L't aI., 1998; Elmol'c et aI., 1963, Mow & Ratcliffe, 1997; SokolofL 1963). Recentlv, experimental measure- ments ha\'c deh:nnined that interstitial fluid pres- surization supports mort' than *YOOk* of the applied load to the canilage surface (Solt/. & Ateshian. 1998) immediately following loading. This ('lfcCl can persist for more than 1,000 scconds and thus shields til(.' EC1VI and chondroc.\'tcs from the crllsh- ing defonnations or the high stresses (20 MPa) re- sulting frollljoilll loading.

CONFINED COMPRESSION EXPLANT LOADING CONFIGURATION

The loading of cartilage in vivo is cxtr('mel~" com- plex. To achic\'C' a better understanding of the de- formational behavior or the tissue under load, an explant loading conliguration known as confined

compression (lvlow et aL, 1980) has been adopted by researchers. In this configuration, a cylindrical car- tilage specimen is fitted snugly il1to a cylindrical, srnZoth-walled (ideally frictionless) confining ring thatprohibits motion and fluid loss in the radial di- rection. Under an axial loading condition via a rigid p~rol1s-permeable loading platen (Fig. 3-9;1), fluid willflo\v from the tissue into the porous-permeable platen, and, as this occurs, ~he cartilage samp~e will compress in creep\_ At any tIme the amount 01 com- pression equals the volume of nuid loss because both the watcr and theECM arc each intrinsicall:v incompressible (Bachrach et aI., 1998). The advan- tage of the confined compression tcst is that it cre- ates a uniaxial, one-dimensional flow and def()rma~ tional J1eld within the tissue, which does not depend on tissue anisotropy.' or properties in the radial diN recLion. This greatly simplir-Ies the mathematics needed to solve the problem.

It should be emphasized that the stress-strain, pressure, fluid, and ion flow fields generated within the tissue during loading can only be calculated; ho\VevcI~ these calculations are of idealized models and testing conditions. There are man:\' confound- ing factors, such as the time~dependentnature and magnitude of loading and alterations in the natural state of prc~strcss (acting \vithin the tissue), that arise from disruption of the collagen network dur~ ing specimen harvesting. Despite limitations in de~ terrnining the natural physiological states of stress and strain within the tissue in vivo, a number of re N searchers have made gains to\\'ard an understand~ ing of potential mechanosignal transduction I11ech~ anisms in cartilage through the use of explant loading studies (Bachrach et aI., 1995; Buschmann et aI., 1992; Kim et aI., 1994; Valhmu et aI., 1998) based on the biphasic constitutive law For soft hy'- drated tissues (Mow et aI., 1980).

BlPHAS1C CREEP RESPONSE OF ARTICULAR CARTILAGE IN COMPRESSION

The biphasic creep response of articular cartilage in a oneNdimensional confined compression ex- periment is depicted in Figure 3-9. **In** this case, a constant compressive stress (To) is applied to the tissue at tinlC to (point A in Fig. 3-98) and the tis~ Slie is allo\ved to creep to its final equilibrium strain (EX). For articular cartilage, as illustrated in the top diagrams, creep is caused by the exuN dation of the interstitial fluid. Exudation is most

rapid initially', as evidenced by the early rapid rate of increased deformation, and it diminishes grad- ually until flow cessation occurs. During creep, the load applied at the surface is balanced by the compressive stress developed within the collagen- PG solid matrix and the frictional drag generated by the flow of the interstitial fluid during exuda- tion. Creep ceases when the compressive stress developed within the solid matrix is sufficient to balance the applied stress alone; at this point no fluid flows and the equilibrium strain EX is reached.

Typically, for relatively thick human and bovine articular cartilages, 2 to 4 mm, it takes 4 to 16 hours to reach creep equilibrium. For rabbit car~ tilage, which is generally less than 1.0 111m thick, it takes approximately' I hour to reach creep equi- librium. Theoretically, it can be shown that the time it takes to reach creep equilibrium varies inversely with the square or the thickness of the tissue (Mow et aI., 1980). Under relatively high loading conditions, > 1.0 J\ilPa, *50 0/()* of the total fluid content may be squeezed From the tissue (Echvards, 1967). Furthermore, in vitro studies demonstrate that if the tissue is immersed in physiological saline, this exuded fluid is fully re- coverable when the load is removed (Elmore et aI., 1963; Sokoloff, 1963).

Because the rate of creep is governed b.y the rate of fluid exudation, it can be used to determine the permeability coefficient of the tissue (Mow et al., 1980, 1989a). This is known as the indirect mea- surement for tissue permeability (k). Average values of normal hun1an, bovine, and canine patellar groove articular cartilage permeability k obtained in this manner arc 2.17 X 10. 15 M·'/N·s, 1.42 x 10'" M'/N·s, and 0.9342 x 10. 15 M4/N·s, respectively (Athanasiou et aI., 1991). At equilibrium, no fluid flow occurs and thus the equilibrium deformation can be used to measure the intrinsic compressive modulus (H,\) 01' the collagen-PG solid matrix (Armstrong & Mow, 1982; Mow et aI., 1980). Average values of normal human, bovine, and canine patellar groove articular cartilage compressive modulus H,.\ are 0.53, 0.47, and 0.55 megapascal (MPa; note 1.0 MPa = 145 Ib/in 2), respectively. Because these coefficients are a measure of the intrinsic material properties of the solid matrix, it is therefore meaningful to determine ho\v they vary \vith matrix composition. It was de- termined that k varies directly, while He' varies in- versely with water content and varies directly with PG content (Mow & Ratcliffe, 1997),

Confining ring

Impermeable platen

"0 1 ~oJw>.~ c rU\_A\_·\_B\_\_•\_\_• • •\_\_•

Time BA

A. A schematic of the confined compression loading configu- ration. A cylindrical tissue specimen is positioned tightly into an impermeable confining ring that does not permit defor- mation (or fluid flow) in the radial direction. Under loading, fluid exudation occurs through the porous platen in the ver- tical direction. **B.** A constant stress *(T"* applied to a sample of articular cartilage *(bottom left)* and creep response of the sample under the constant applied stress *(bottom right).* The

No exudation C TimeEquilibrium (C) 6Copious fluid exudation

Creep (6) Unloaded (A)

oa.~ E-' 0;oEao 0-w:e>- - =-..........\_----...-i

Equilibrium deformation 1

drawings of a block of tissue above the curves illustrate that creep is accompanied by copious exudation of fluid from the sample and that the rate of exudation decreases over time from points A to B to C. At equilibrium (EO'-'), fluid flow ceases and the load is borne entirely by the solid matrix (point C). *Adapted from* MOH~ VC, *Kuei,* S,C, *Lai, Wl'vl., f:.'( af. (1980). Biphasic creep and stress relaxation of articular cartilage* in *com- pression: Theory and experiments.* J Siomech Eng, 102, 73-8fl