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Orthotic Prescription

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KEY POINTS

- Orthoses provide necessary biomechanical stability to support musculoskeletal structures, facilitate healing, and improve patient participation in activities of daily living.
- Recent advances in technology and materials have increased options for orthotic design and intervention.
- Orthotists use information related to the patient's pathology and prognosis, biomechanics, muscle strength, joint range of motion, environmental factors, psychosocial situation, and goals to design a device that meets the prescribing provider's goals.
- Prescribing providers must include sufficient information related to the patient and the needs for orthotic care in orthotic prescriptions. Certain information must be supplied by the prescribing provider and included in the patient's medical record to ensure third-party payer coverage of orthotic care.

INTRODUCTION

Orthoses can be an integral part of a rehabilitative plan of care through their external support of musculoskeletal structures and function. Orthotic care and technology have advanced tremendously in the past half-century,[1,2] and these advancements have facilitated new orthotic designs and device evolution.[1] The foundational principles of force systems, biomechanics, and materials science continue to inform orthotic decision-making, with newer materials and manufacturing processes allowing for improvements in design and care.[1] Orthotic recommendations are informed by the patient's medical condition and prognosis, the health care team's rehabilitation goals, and patient-specific factors such as biomechanics, muscle strength, joint range of motion, and activities of daily living.

Recognizing these advances in orthotic design and the need for interdisciplinary and team-based care, entry-level orthotic and prosthetic clinical education must now be obtained at the graduate level.[3,4] Orthotists receive education, clinical training, and continuing professional development, which allows them to make informed and evidence-based design decisions under the referral of a prescribing provider.[4-6] Importantly, prescribing providers can facilitate interprofessional communication and limit barriers to patient care by writing a clear prescription, providing necessary patient information, including the need for and anticipated functional benefit of the device, and consulting with the orthotist when needed.

NAMING AND CLASSIFICATION OF ORTHOSES

The International Organization for Standardization (ISO) defines orthoses as externally applied devices used to compensate for impairments of the structure and function of the neuromuscular and skeletal systems.[7] The ISO has named orthoses by the joints they cross, and acronyms are commonly used in place of full text (Table 1.1). Joints are listed in order from distal to proximal or superior to inferior. For example, an orthosis that crosses the ankle is called an ankle-foot orthosis, or AFO (Fig. 1.1). An orthosis that crosses the hand, wrist, and elbow is called an elbow-wrist-hand orthosis, or EWH. An orthosis that crosses the thoracic, lumbar, and sacral vertebrae is called a thoracolumbosacral orthosis, or TLSO (Fig. 1.2).

There are cases in which a specific orthosis may be named by the manufacturer or original inventor of the device, such as the Arizona AFO[8] and Jewett spinal orthosis.[9] In some cases, these devices are now fabricated by a variety of companies; in others, design and fabrication may only be available through limited providers. While some orthotic devices are available in a variety of sizes over the counter or off the shelf, most are custom-fit or custom fabricated. Custom-fit devices are prefabricated devices that require some modification or adjustment to properly fit the patient. Custom-fabricated devices are individually made for a specific patient using measurements, impressions, or scans of the body.[10]

ORTHOTIC DECISION-MAKING AND DESIGN

The International Classification of Function (ICF)[11] outlined by the World Health Organization offers a framework through which a provider can make a multidimensional recommendation for orthotic care. The ICF centers around the activities of the patient and encourages providers to consider influential domains of health condition, body structure and function, participation, and environmental and personal factors. Researchers have documented that clinicians primarily use assessments of body structure and function and participation when determining orthotic intervention.[12] Range of motion, muscle strength, gait patterns, and activities of daily living are key components of an assessment for orthotic intervention. However, clinicians should also consider the other domains of the ICF when determining the most appropriate device for an individual. Factors such as the ability to independently don and doff an orthosis (body function, personal factors), the appearance of the orthosis (personal factors, environmental factors), and disease progression (health condition) influence the wearer's experience with the device. These domains may be the difference between device acceptance and device abandonment. For example, a short-term device to address an acute diagnosis may be designed

TABLE 1.1 Summary of Orthotic Naming Conventions and Acronyms

Name of Orthosis and Joints Crossed	Acronym
Finger orthosis	FO[a]
Hand orthosis	HO
Wrist-hand orthosis	WHO
Elbow-wrist-hand orthosis	EWHO
Shoulder orthosis	SO
Cervical orthosis	CO
Cervicothoracic orthosis	CTO
Thoracolumbosacral orthosis	TLSO
Lumbosacral orthosis	LSO
Foot orthosis[a]	FO[a]
Ankle-foot orthosis	AFO
Knee-ankle-foot orthosis	KAFO
Hip-knee-ankle-foot orthosis	HKAFO

[a] The provider should use the patient's diagnosis as context when interpreting 'FO.' The acronym 'FO' may be used for finger orthosis or foot orthosis.

Figure 1.1 A solid ankle-foot orthosis with a molded inner boot.

Figure 1.2 A thoracolumbosacral orthosis designed to treat scoliosis.

very differently than a long-term device designed for chronic conditions. While the biomechanical goals may be similar, design and material choices may need to be adjusted so that the patient and the device can sustain long-term force application. Validated outcome measures such as the Orthotic and Prosthetic Users Survey,[13,14] the Activities and Balance Confidence Scale,[15] and the Timed Up and Go (TUG) test[16] aid the clinician in decision-making, post-fit assessment, and follow-up. A plan for follow-up and reevaluation of the patient's status is critical to ensuring the appropriate use or discontinuation of use of an orthosis. The use of self-reported and clinically assessed outcome measures provides an opportunity for longitudinal assessment of the appropriateness, function, and impact of an orthotic device.

Consideration of the needs at the affected body segment and the effects on proximal body segments is critical to decision-making and device effectiveness. An orthosis directly impacts the body segment it covers and indirectly impacts more proximal or adjacent segments.[17] Limiting motion at one joint may increase compensatory motion at another joint; providing stability at one joint may improve biomechanical function of another. Alternatively, preventing motion at one joint may have negative effects on more proximal segments.[18] An orthotic prescription, however, may require weighing the immediate needs of one segment over the consequences for other body segments (e.g., a spinal orthosis needed to facilitate healing of a compression fracture). Conversely, the design of a long-term orthosis may be informed by broader segmental needs, such as the impact of an AFO design on knee stability. In most cases, biomechanical needs and goals, as discussed by members of the rehabilitation team, generate the initial orthotic design considerations.

In addition to reviewing the provider's prescription and any associated health records, orthotists review the patient history, conduct a physical exam, and analyze the patient's gait to inform decisions and orthotic design. The orthotist may begin an evaluation by taking a medical history and observing posture, gait, or movement patterns. Use of video recording and scoring tools, such as the Edinburgh Visual Gait Score,[19] may be used to quantify and document findings. An assessment of sensation, joint range of motion, and muscle strength (through manual muscle testing)[20] informs choices for orthotic trimlines, articulation, joints, and materials. Moreover, the assessment of muscle length may contribute to decisions on joint positioning. For example, the length of the iliopsoas muscle may determine the position of an orthotic hip joint, and an appropriately performed modified Thomas test[21] (Fig. 1.3) may provide necessary information on the presence of a contracture. The length of the hamstring muscles, which can be assessed through measuring the popliteal angle[22] (Fig. 1.4), may impact lower limb orthotic alignment at the knee joint or in relation to the inclination of the AFO. Similarly, the length of the gastrocnemius muscle may determine the angle of an AFO. Orthotists must properly

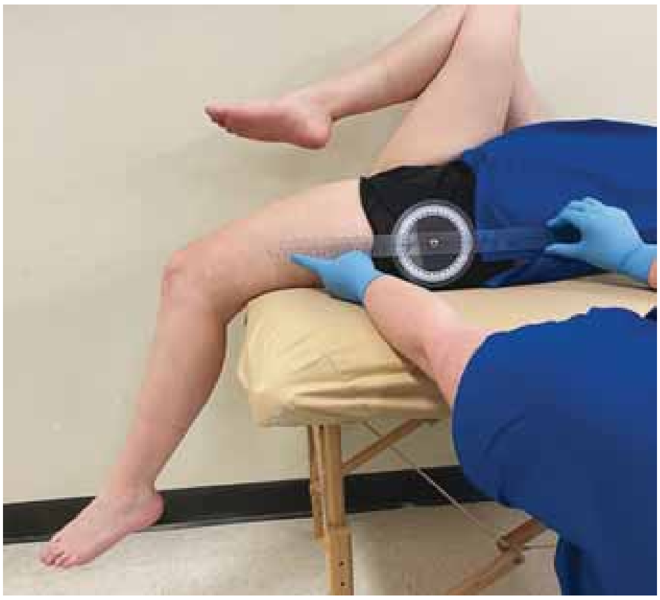

Figure 1.3 Use of the Thomas test to evaluate for iliopsoas contracture.

Figure 1.4 Use of a popliteal test to evaluate for hamstring contracture.

perform a Silfverskiold test (Fig. 1.5) to assess for the presence of equinus and determine gastrocnemius length.[23]

Assessment of bony torsion and ligamentous laxity can also play a critical role in orthotic design. Femoral and tibial torsion will affect the alignment of hip, knee, and ankle joints. Overlooking the alignment of these joints may result in device malfunction and, ultimately, failure. An orthotist may use Craig's test to evaluate femoral torsion (Fig. 1.6), as femoral torsion and resulting lever-arm dysfunction have been shown to cause pathologic loading of the lower limb.[24] Evaluating the rotational profile of a limb is critical in individuals with chronic neuromuscular conditions.[25] Measurement of tibial torsion through an assessment of the thigh-foot angle (Fig. 1.7), which has demonstrated reliability and consistency with imaging,[26] is necessary to appropriately design a knee-ankle-foot orthosis (KAFO). Verifying the integrity of joint ligaments, such as medial, lateral, and cruciate ligaments, informs orthotic design as well. In cases of ligamentous instability, an orthosis may extend across a joint line to provide enhanced stability.

Figure 1.5 Ankle joint dorsiflexion measurement with the knee in extension and the midfoot in a neutral position.

Figure 1.6 Assessing femoral anteversion using Craig's test.

Figure 1.7 Assessing tibial torsion using the thigh-foot angle.

A thorough examination in the context of orthotic intervention should also include an assessment of sensation and muscle tone, when appropriate. In cases of spinal cord injury, for example, an evaluation of dermatomes and myotomes aids the orthotist in both decision-making and educating the patient on the importance of daily skin checks to immediately address skin breakdown. Determination of muscle hypertonicity using a Modified Ashworth Scale[27] may aid in determining whether to proceed with a solid or articulated device and provide important baseline information for comparison at a follow-up appointment. Conversely, observation of hypotonicity may indicate a need for additional support to aid recovery or development, such as in cases of flatfoot associated with Down syndrome.[28] Information related to sensation and muscle tone is critical in the acute rehabilitation phase and influences the plan of care when changes are documented and correspond to needs for revisiting the orthotic intervention.

Over half of orthotic practice involves the lower limb,[29] and orthotic intervention may facilitate the return of ambulation and activities of daily living. Orthotic intervention may also be used to facilitate normal bone growth and prevent the progression of bony deformation, such as in plagiocephaly,[30] Blount disease,[31] or scoliosis.[32] Orthoses may also be used to address soft tissue contractures or gait abnormalities[33] and facilitate post-surgery tissue healing.[34] Limiting, resisting, and assisting joint motion are mechanisms through which orthoses may contribute to a plan of care. Moreover, circumferential compression may be used to stabilize bony tissue[35] or increase cavitary pressure to reduce the load placed on an injured spine.[36]

When the prescribing provider provides primary biomechanical goals, either through medical documentation or detailed prescriptions, the orthotist can ensure the appropriate design of an orthosis. For example, if a provider indicates a need to maintain the ankle in a fixed degree of plantarflexion following surgical repair of the Achilles tendon, an orthotist can provide a device that meets those needs, as well as any anthropometric or personal needs of the patient. Similarly, if a prescribing provider identifies a need for an orthosis to improve gait biomechanics, an orthotist can perform an evaluation and assessment to design a lower limb orthosis that meets those goals. While prescribing providers are encouraged to provide as detailed orthotic recommendations as possible, it is appropriate and beneficial to consult with an orthotist if there are questions related to device specificity or efficacy.

The goals of the health care team are considered in conjunction with patient-centered aspects of orthotic decision-making. A patient's activities of daily living, vocation, and avocation are all considered by the orthotist when designing the orthotic intervention. Cosmesis is often a concern for patients who need an orthotic device. The orthotist must consider the preferences of the patient, when possible, without sacrificing the goals of orthotic intervention; patients are often able to choose device colors or patterns, particularly if the device is custom made. An orthotist selects materials within capacity of the patient's weight and activity level and in alignment with the goals for orthotic intervention. These materials traditionally included metal (aluminum and stainless steel) and leather but now include thermoformable foams and plastics, composites, and additive manufacturing materials. When a patient's medical condition is expected to progress or improve over time, the orthotist may also schedule follow-up appointments to evaluate the efficacy of the orthosis. Continued communication between the orthotist and the prescribing provider is essential in such instances.

The fabrication process begins when the orthotist takes an impression or a scan of the segment of the body necessary to fabricate the orthotic device. The impression is called a negative mold. For example, for fabrication of an AFO, an impression is taken encompassing the lower extremity, distal to the knee. Impressions can be made with plaster or fiberglass bandage. A positive mold is then created by inverting the cast of a negative mold by filling it with plaster, alginate, or similar materials that will take the shape of the cast segment. This impression is called a positive mold. Another option for impression is a digital scan of the limb. This method yields a direct digital positive mold. The positive mold is modified to form the shape of the orthotic device. The orthotist selects the appropriate thermoplastic or lamination materials to create a final device from the positive mold. A final device is thermoformed or laminated around the mold, or according to measurements, or produced via additive manufacturing methods.

DOCUMENTATION AND PRESCRIPTION REQUIREMENTS

While a simple prescription is sufficient to initiate orthotic care, a detailed written prescription is often necessary for health insurance approval and coverage. Currently, third-party payers require documentation in the prescribing provider's notes, in addition to the orthotist's notes, that supports medical necessity for orthotic intervention.[37] As such, prescribing providers must be familiar with policy articles pertaining to orthotic devices[21] (Table 1.2). These resources equip providers with detailed information according to which devices are considered medically necessary for specific diagnoses and regarding which criteria must be met to receive insurance coverage for those devices. Documentation of these criteria must be provided in the prescribing provider's notes for insurance coverage to be obtained. Most, but not all, private insurance companies follow Medicare and Medicaid guidelines. Orthotists and their administrative staff are intimately familiar with these requirements and often ask for documentation to support them. Communication between the prescribing provider and orthotist regarding these needs ahead of a patient referral can often reduce wait time for insurance prior authorization or approval. Many times, a letter of medical necessity from the prescribing provider is needed to clearly document the indications for orthotic intervention, the need for this intervention to support the patient's health status and function, and the anticipated risks should the device not be supplied to the patient.

TABLE 1.2 Insurance Coverage Criteria[a]

Policy Article	Coverage Indication Overview[b]
Ankle-foot orthoses (AFOs) not used for ambulation	1. Plantarflexion contracture of the ankle with dorsiflexion on passive range of motion testing of at least 10 degrees (i.e., a nonfixed contracture); and 2. Reasonable expectation of the ability to correct the contracture; and 3. Contracture is interfering or expected to interfere significantly with the beneficiary's functional abilities; and 4. Used as a component of a therapy program that includes active stretching of the involved muscles and/or tendons; and 5. The beneficiary has plantar fasciitis.
Custom knee orthoses (KOs)	1. Deformity of the leg or knee 2. Size of thigh and calf 3. Minimal muscle mass upon which to suspend an orthosis
AFOs and knee-ankle-foot orthoses (KAFOs) used for ambulation	1. Require stabilization for medical reasons, and 2. Have the potential to benefit functionally
Custom AFOs and KAFOs used for ambulation	1. The beneficiary could not be fit with a prefabricated AFO; or 2. The condition necessitating the orthosis is expected to be permanent or of longstanding duration (more than 6 months); or 3. There is a need to control the knee, ankle, or foot in more than one plane; or 4. The beneficiary has a documented neurologic, circulatory, or orthopedic status that requires custom fabricating over a model to prevent tissue injury; or 5. The beneficiary has a healing fracture that lacks normal anatomical integrity or anthropometric proportions.
Spinal orthoses: Thoracolumbosacral orthoses (TLSOs) and lumbosacral orthosis (LSOs)	1. To reduce pain by restricting mobility of the trunk; or 2. To facilitate healing following an injury to the spine or related soft tissues; or 3. To facilitate healing following a surgical procedure on the spine or related soft tissue; or 4. To otherwise support weak spinal muscles and/or a deformed spine.

[a]An overview of basic criteria that must be met and documented in the prescribing provider's notes for the Centers for Medicare and Medicaid Services to provide coverage for certain orthotic devices. This information does not include specific information related to diagnosis codes, device codes, and certain orthotic components.

[b]Refer to Centers for Medicaid and Medicare Services (CMS) for comprehensive information, including the International Classification of Diseases (ICD-10) coding, the Healthcare Common Procedure Coding System (HCPCS), and additional criteria for specific components.

The written order, which initiates orthotic treatment, must often be followed by a detailed written order in cases in which a third-party payer is involved. This detailed order is generated by the orthotist and includes codes specific to the design and components of the orthosis. This documentation requires approval from the prescribing provider; in cases in which an orthotist's evaluation and recommendation deviate from the written order, communication between members of the health care team is crucial to ensure consistent and appropriate plans of care. Orthotists must communicate with the prescribing provider when seeking consideration for an alternative orthotic design or approach than what was described on the written order.

A thorough evaluation of a patient who requires orthotic care will include a medical history, discussion of environmental contexts, identification of diagnosis-specific or plan-specific orthotic needs, assessment of neuromuscular and sensory function, and gait analysis. Inclusion of the patient's goals and concerns in the decision-making process will improve outcomes. Each component of an orthotic recommendation requires an associated patient-specific biomechanical or functional justification. Inclusion of such information across the health care team's records will facilitate interdisciplinary communication and support medical necessity.

INTERDISCIPLINARY AND TEAM-BASED CARE

Clear and open communication between an orthotist and a prescribing provider is paramount to ensuring comprehensive and timely patient care. A provider's prescription must include the patient's name, diagnosis requiring orthotic treatment, relevant additional diagnoses, date of encounter with the prescribing physician, and estimated length of time orthotic care will be needed. Communication of this information and relevant medical records across the health care team facilitates improved patient outcomes, and patients who receive orthotic care will benefit from interdisciplinary communication and decision-making. In many cases, a prescribing provider, orthotist, and physical or occupational therapist will be involved in the plan of care for a patient receiving orthotic treatment. Orthotists are trained to be experts in their professional domain and rely on the expertise of, and communication from, the prescribing providers and therapists to best contribute to positive patient outcomes. Advances in orthotic materials, design, and technology will continue to grow in the next decades, making it increasingly difficult for all members of the health care team to be aware of all the orthotic options available. As a result, the team's communication with the orthotist will play an even greater role in determining the most appropriate orthotic plan of care.

CASE STUDY

The orthotist will receive an orthotic prescription from a physician. Physician prescriptions vary in the amount of detailed information about the device ordered. It is within the orthotist's scope of practice to provide a full orthotic recommendation in response to the prescription generated by the physician. This recommendation includes all components of the device, with clear justification on how these components will address the patient's needs.

Case Details

A 13-year-old male patient presents to a clinic with a prescription for lower limb orthotic intervention. He has a primary diagnosis of lower lumbar spina bifida and a secondary diagnosis of lower limb weakness. He reports that he has difficulty keeping up with his friends at school and would like his devices to help stabilize him when he walks, allowing him to walk for longer distances without fatigue. He uses forearm crutches to combat the fatigue when out in the community, but he can navigate his home without them most of the time. He reports that his previous devices are now too small and have caused skin breakdown on his feet; as a result, he is hesitant to use them. His mother reports she is concerned about the increased difficulty he has with walking, and she wants to prevent worsening of his alignment. The patient lives in a single-story home with his mom. He can don and doff his previous devices independently. He reports a history of two falls within the past month that did not result in injury.

Select Physical Exam Findings

- Postural analysis and observation:
  - Weight-bearing lower limb posture demonstrates increased hip, knee, and ankle flexion.
  - Mild hip adduction and the appearance of genu valgum are present bilaterally.
  - Observation of internal rotation of the femur and tibia, accompanied by hindfoot valgus, midfoot collapse, and forefoot abduction.
  - Erythema is present on the navicular tuberosity bilaterally.
- Observational gait analysis: The patient is 5 feet, 2 inches tall and weighs 145 lb. Observational gait analysis demonstrates foot-flat initial contact; excessive knee flexion at loading response; knee extension thrust, hindfoot valgus, and midfoot pronation at early midstance; delayed heel rise (excessive ankle dorsiflexion) at terminal stance; and bilateral Trendelenburg gait (ipsilateral trunk lean). During the swing phase, the patient demonstrates posterior trunk lean.
- Assessment of sensation: Limited sensation on the dorsal and lateral aspect of the foot and ankle, bilaterally.
- Assessment of joint range of motion and foot flexibility:
  - Full passive bilateral knee joint range of motion.
  - Full passive bilateral ankle joint dorsiflexion range of motion.
  - Passive ankle joint plantarflexion to 20 degrees.
  - When the knee is in full extension, the hindfoot, midfoot, and forefoot are correctible to a neutral position up to 5 degrees of plantarflexion bilaterally.
- Assessment of muscle tone: Lower limb presents as normal on the Modified Ashworth Scale; hypotonicity observed in the foot and ankle.
- Assessment of muscle length:
  - Hamstring length (popliteal angle): 5 degrees of flexion bilaterally.
  - Silfverskiold test: 5 degrees of plantarflexion bilaterally.
- Assessment of muscle strength (five-point scale):
  - Hip: Flexors 5/5, extensors 3/5, abductors 3/5
  - Knee: Flexors 4/5, extensors 4/5
  - Ankle: Dorsiflexors 2/5, plantarflexors 2/5
- Ligamentous testing of the knee: Negative varus and valgus stress tests; negative anterior and posterior drawer tests.
- Thigh-foot angle: 10 degrees
- TUG test (with forearm crutches): 9 seconds
- Orthotic and Prosthetic Users Survey, Lower Extremity Functional Status: 52/80

Rationale for Orthotic Prescription and Treatment Plan

Custom AFO rationale: An AFO is indicated because the patient (1) requires stabilization for medical reasons and (2) has the potential to benefit functionally from the device. His lower limb alignment (genu valgum, torsion, and triplanar deformity) indicates the need for a custom device. Moreover, his condition is of longstanding duration, there is a need to control the limb on more than one plane, and he has a documented neurologic status (sensory deficits) that requires custom fabrication to prevent injury. There are no indications that this patient requires an orthotic device that crosses the knee joint.

AFO design considerations: A total-contact design around the foot and ankle that accommodates plantarflexion contracture and maintains the hindfoot, midfoot, and forefoot in a neutral position will aid in triplanar control and prevent skin breakdown (force application over a larger area). An AFO that resists or blocks dorsiflexion will aid with restoring the knee extension-plantarflexion required for terminal stance. An AFO that resists plantarflexion will aid in swing limb clearance. An AFO that provides a dynamic component (through a carbon fiber posterior strut, for example) may improve the patient's stamina. The ankle angle of the AFO will need to accommodate the plantarflexion contracture. Alignment of the AFO and the AFO-footwear combination must address the knee extension thrust, plantarflexion contracture, and knee flexion contracture. Overall, the material choices require the device to be as lightweight as possible while providing sufficient stability and control to prevent falls and maintain upright posture and skeletal alignment.

Follow-up: The patient will require a series of follow-up appointments to ensure appropriate fit and function of the AFO.

Physical therapy: The patient would benefit from physical therapy to address the lower limb muscle contractures and joint range of motion limitations. Once fit with the new AFOs, a series of physical therapy appointments focusing on gait training and optimization of the orthoses, ideally in concert with the orthotist, will provide the most optimal outcomes for the patient.

Education: because of the patient's sensory deficits in the lower limb, education on proper skin checks is paramount. The orthotist should also educate the patient and his mother on a break-in schedule, care and maintenance of the orthosis, and when to schedule follow-up appointments.

The full reference list for this chapter is available in our eBook – see inside front cover for access details.

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2

Materials Science

Jared A. Howell

KEY POINTS

- One of the fundamental considerations for a prosthetist or orthotist is the material, make up, and structural integrity of the orthosis or prosthesis they will be providing to their patients. A fundamental understanding of materials science ensures that the orthosis or prosthesis performs the intended purposes while remaining safe to use, efficient, and aesthetically pleasing. When principles of materials science are understood and applied effectively, providers can innovate, customize, and improve both form and function.
- While not a comprehensive summary of materials science, this chapter is designed to introduce the reader to the underlying principles of materials science as they pertain to the development and manufacture of orthoses and prostheses. Specific attention is given to each of the following areas:
  - Material properties: Understanding the unique features and attributes of any given material as they apply to its use within the orthotic and prosthetic (O&P) profession.
  - Stress and strain: The scientific study of how materials behave when placed under load. This includes an in-depth understanding of different load types and how materials respond to the application of different loading mechanisms.
  - Deformation and failure: The causes and effects of elastic and plastic deformation, yield, and ultimate strength. This includes other modes and mechanisms of failure, including creep and fatigue failure.
  - Material selection: The scientific approach by which materials are compared against one another and even optimized for a specific application. Particular emphasis is placed on biomechanical principles and the application of those loads to the materials restraining them.
  - Material processing: All materials used in O&P require some processing to achieve the desired shape or profile.

Great advancements in materials have occurred in the last century; of particular note is the move from refined raw materials used in their pure form to advanced materials developed through innovative chemistry, which includes the combination of multiple elements to optimize specific material properties. New materials are being developed regularly, and their application in the practice of orthotics and prosthetics (O&P) has led to added comfort in the skin/device interface, improved strength-to-weight ratios in finished devices, enhanced fit and anatomic congruence, the advent of tailored dynamic elements through structured deflection, improved profiles and cosmesis, and an added layer of precision that was not previously possible.

UNDERSTANDING MATERIALS SCIENCE TO MAXIMIZE PATIENT SAFETY

The fabrication of orthoses, prostheses, and other assistive devices almost always involves the use of combinations of different materials. Fig. 2.1 shows an example of thermoplastic and thermoset material being used together to provide better strength properties to both. Furthermore, there is typically a combination of prefabricated and custom-made components made from different materials with unique material properties. An understanding of each material's primary properties, performance characteristics, and ability to withstand or counteract external load can ensure that the patient or user of the orthosis or prosthesis remains safe and that the desired biomechanical outcome can be achieved.

Acknowledgements: We would like to acknowledge the extensive work of the prior edition authors, Thomas R. Lunsford and Bill Contoyannis.

Orthoses and prostheses are often fabricated from a variety of different materials with significantly different properties, including metals, plastics, leathers, composites, foams, and synthetic rubbers. The optimized assistive device makes use of each of the material properties to achieve a specific function, some structural and others protective or accommodative. An education in materials science helps ensure the clinical parameters and engineering strength are adequately balanced to achieve the desired result, which is to restore mobility and function.

Selection of the correct material for a given design depends partially on understanding the elementary principles of mechanics and materials; concepts of forces; deformation and failure of structures under load; improvement in mechanical properties by heat treatment, work (strain), hardening, or other means; and design of structures. For example, available choices for the components of a knee-ankle-foot orthosis (KAFO) may include different steels, aluminum, and titanium with their respective alloys. Important but minor uses of other metals include more malleable fasteners such as steel, copper, or brass rivets or different sizes and styles of steel screws. Plastics, fabrics, synthetic rubbers, and leathers have wide indications and are frequently used in skin-contact applications. Composite structures (polymer matrices with reinforcing materials) have widespread application in modern orthoses and prostheses. The most common of these composites are carbon fiber or glass reinforced polymers. Some of the primary clinical challenges come when materials with far different structural properties are combined to meet the needs of the patient. A simple example is the attachment of a flexible Dacron strap to a thermoplastic ankle-foot orthosis (AFO) using a thin steel or copper rivet.

Given the complexity and variability of the human body and the condition that requires treatment, it is unwise to assume that any single

Figure 2.1 A transparent diagnostic socket reinforced using preimpregnated carbon fiber. Although this is not structurally a desirable solution, it may be clinically necessary.

material is a panacea; it is also safe to assume that an orthosis or prosthesis may benefit from having divergent mechanical properties. Consider a preimpregnated (prepreg) carbon fiber orthosis designed to be rigid and thin combined with a metal or polymer joint so the ankle can articulate during the gait cycle. The fiber reinforced polymer provides structural stability throughout portions of the gait cycle while the joint is designed to articulate and move in select directions to accommodate ankle movement and stability. The metal joints are wear resistant and designed to withstand millions of cyclical movements. In addition, practitioners are rarely presented with situations that require only one material or with single-design situations that do not require modification, customization, or variation over time. Despite the addition of materials such as prepreg carbon fiber and some newer materials created to meet the growing demand for 3D printing (additive manufacturing), the basic materials science discussed in this chapter remains largely unchanged.

In general, the practitioner's understanding of the mechanics and strengths of materials, even if intuitive, is important during the design stage. A general understanding of stresses arising from loading of structures, particularly from the bending of beams, is needed. The practitioner can then appreciate the importance of simple methods that allow controlled deformation during fitting, provide stiffness or resiliency as prescribed, and reduce breakage from impact or repeated loading. A general discussion of materials and specific theory related to design, fabrication, riveting guidelines, troubleshooting, and failure considerations follows.

Consideration should be given to the international standards of terminology that are used to describe orthotics, prosthetics, properties of materials, and units of measure (whether imperial or metric) and the engineering principles used to describe the various effects of loading upon these materials.

IMPERIAL AND METRIC CONVERSIONS

Most of the examples provided here are presented using both imperial and metric units because both are still used with frequency when defining and comparing material properties. Some examples will assist with the general "comparison" between imperial and metric units.

- 1 pound (lb) = 0.45 kilograms (kg)
- 1 kilogram (kg) = 9.8 Newtons (N) of force (the same as 1 kg × gravity, or 9.8 meters per second)
- 1 inch (in) = 0.025 meters (m) or 2.5 centimeters (cm)
- 1 meter (m) = 39.3 inches (in)
- 1 meter (m) = 100 centimeters (cm)
- 1 centimeter (cm) = 10 millimeters (mm)
- 1 pound per square inch (psi) = 6895 Pascals (Pa) (or 0.006895 megapascals [MPa; or 1 million pascals])
- 1 Pascal (Pa) = 1 Newton per square meter (N/m^2)
- Stress units: pounds per square inch or megapascals (million Newtons per square meter)
- Strain units: fraction of an inch per inch or fraction of a meter per meter

STRENGTH AND STRESS

One of the practitioner's main considerations is the strength of the material selected for fabrication of orthoses or prostheses. *Strength* is defined as the ability of a material to resist forces. When comparative studies are made of the strengths of materials, the concept of stress is critical to understanding and categorizing a material's expected performance.

Stress relates to both the magnitude of the applied forces and the amount of the material's internal resistance to the forces. *Stress* is defined as force per unit cross-sectional area of material and is usually expressed in pounds per square inch (imperial) or pascals or megapascals (metric). The amount of stress (σ) is computed using the equation:

$$\sigma = \frac{F}{A} \quad \textbf{(2.1)}$$

where F = applied force (pounds or Newtons) and A = cross-sectional area (square inches or square meters).

The same amount of force applied over different areas causes radically different stresses. For example, a 1-lb weight (about 0.5 kg or 4.9 N) is placed on a cylindrical test bar with a cross-sectional area of 1 in^2 (about 6.5 cm^2). According to Eq. 2.1, the compressive stress σ_c in the cylindrical test bar is 1 lb/in^2 or about 7538 Pa (Fig. 2.2). When

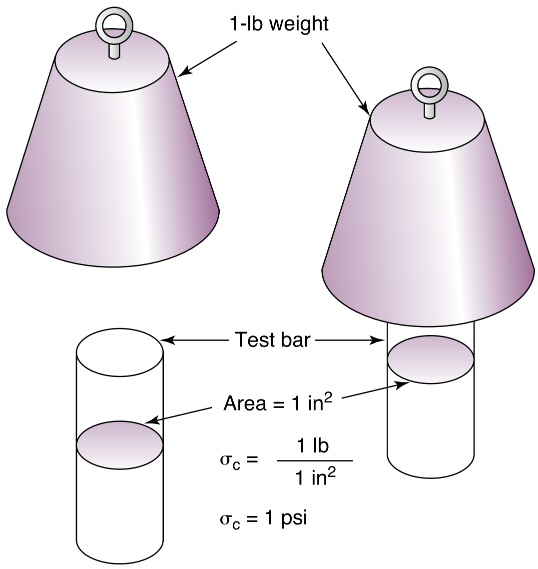


Figure 2.2 Compressive stress on a cylinder. *psi,* Pounds per square inch.

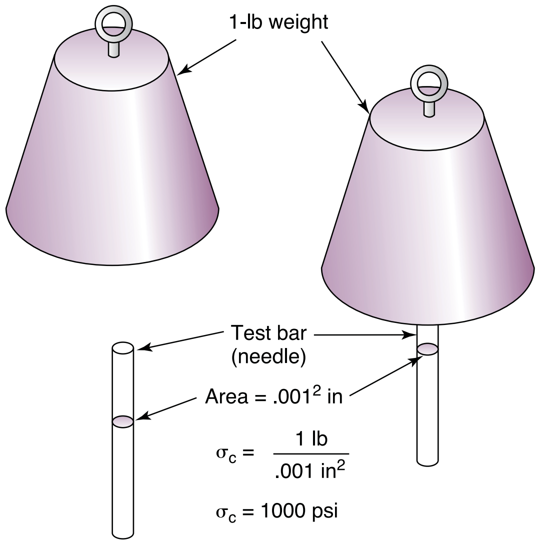


Figure 2.3 Compressive stress on a needle. *psi,* Pounds per square inch.

the same 1-lb weight is placed on a needle with a cross-sectional area of 0.001 in^2 (0.0065 cm^2), the compressive stress σ_c in the needle is 1000 psi or 7.5 MPa (or 7,5000,000 Pa) (Fig. 2.3).

A force exerted on a small area always causes more stress than the same force acting on a larger area. When a woman wears high-heeled shoes, her weight is supported by the narrow heels, which have an area of only a fraction of a square inch. With flat shoes, the same weight or force is spread over a heel with a larger cross-sectional area. The stress in the heel of the shoe is much greater when high-heeled shoes are worn because less material is resisting the applied forces. As an orthotist or prosthetist, this principle is fundamental and should always be considered; it may be better to increase the surface area withstanding the force rather than choose a more expensive or exotic material with greater strength. Good design requires optimization of both parameters.

Similar problems are encountered in orthoses and prostheses. A child who weighs 100 lb (45 kg) wearing a weight-bearing orthosis with a 90-degree posterior stop (Fig. 2.4) can exert forces at initial contact that create stresses of thousands of pounds per square inch. If the child jumps, this can increase the forces imparted by three to five times the body weight of the child. The ability of a material or component to withstand failure is directly related to the stress applied and the ability of the component or material to withstand it. Assuming a consistent force is applied, the overall stress on the material is reduced dramatically by increasing the surface area of the applied load, even if no other changes are made. Engineering stresses for each material are calculated using a prescribed testing procedure governed by accepted international standards. For most materials, peak stress values are shown on material properties charts. Examples of these charts with relevant O&P materials are represented later in the chapter. It is also important to note that the fabrication techniques used in O&P may not be sufficiently controlled to ensure the materials selected actually perform at the ratings shown in the charts provided. These numbers should act as guides, and the comparison of different strength values can be particularly helpful when examining alternative material options with improved properties.

Tensile, Compressive, Shear, and Flexural Stresses

Materials are subject to several types of stresses depending on the way the forces are applied: tensile, compressive, shear, and flexural.

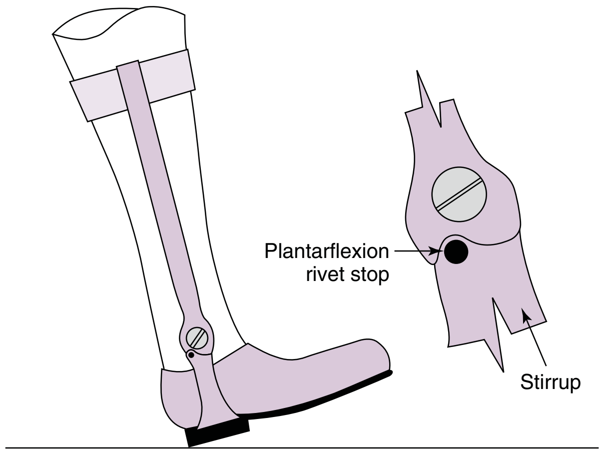


Figure 2.4 Ankle-foot orthosis with 90-degree plantarflexion stop.

Tensile Stresses

Tensile stresses directly pull apart an object or cause it to be in *tension.* Tensile stresses occur parallel to the line of force but perpendicular to the area in question (Fig. 2.5). If an object is pulled at both ends, it is in tension, and sufficient force will pull it apart. Two children fighting over a fish scale and exerting opposing forces put it in tension, as shown by the indicator on the scale (Fig. 2.6). Strings or ropes are good examples of objects that typically can only have tension applied, as they do not provide any resistance in compression. Per example: Raw carbon fiber strands are only effective if loaded in tension.

Compressive Stresses

Compressive stresses act to squeeze or compress objects. They also occur parallel to the line of force and perpendicular to the cross-sectional area (Fig. 2.7).

Many materials may be strong in compression and relatively weak in tension. The opposite can also be true. As a corollary to the fiber or

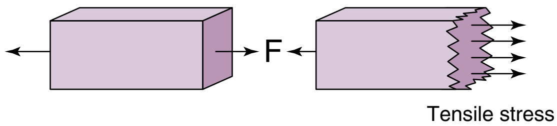


Figure 2.5 Tension. *F,* Force.

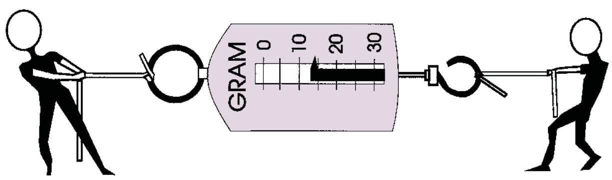


Figure 2.6 Spring scale used to demonstrate tension.

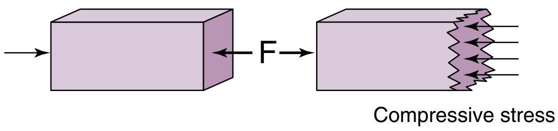


Figure 2.7 Compression. *F,* Force.

rope example referenced above for tension, foams are widely used to distribute compressive forces inside a prosthesis or orthoses but are completely ineffective at resisting tensile loads.

Shear Stresses

Shear stresses act to scissor or shear the object, causing the planes of the material to slide over each other. Shear stresses occur parallel to the applied forces. Consider two blocks (Fig. 2.8A) with their surfaces bonded together. If forces acting in opposite directions are applied to these blocks, they tend to slide over each other. If these forces are great enough, the bond between the blocks will break (Fig. 2.8B). If the area of the bonded surfaces were increased, however, the effect of the forces would be distributed over a greater area. The average stress would be decreased, and there would be increased resistance to shear stress.

Common lap and clevis joints are examples of a shear pin used as the axis of the joint (Fig. 2.9). The lap joint has one shear area of the rivet resisting the forces applied to the lap joint (Fig. 2.9A), and the rivet in the box joint (clevis) has an area resisting the applied forces that is twice as great as the area in the lap joint (assuming that the rivets in both joints are the same size; Fig. 2.9B). Consequently, the clevis joint will withstand twice as much shear force as the lap joint. The lap joint also has less resistance to fatigue (fluctuating stress of relatively low magnitude, which results in failure), because it is more susceptible to flexing stresses.

Flexural Stress

Flexural stress (bending) is a combination of tension and compression stresses. Beams are subject to flexural stresses. When a beam is loaded transversely, it will sag. The top fibers of a beam are in maximum compression while the bottom side is in maximum tension (Fig. 2.10). The term *fiber*, as used here, means the geometric lines that compose the prismatic beam. The exact nature of these compressive and tensile stresses is discussed later.

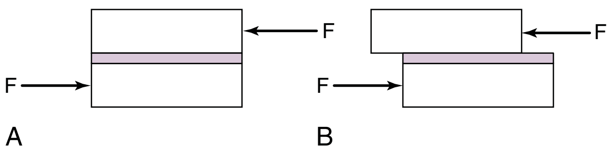


Figure 2.8 (A and B) Shear. *F*, Force.

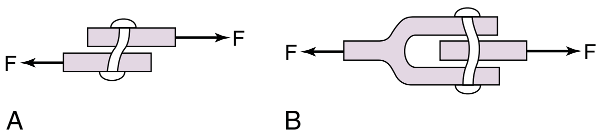


Figure 2.9 (A and B) Joint shear. *F*, Force.

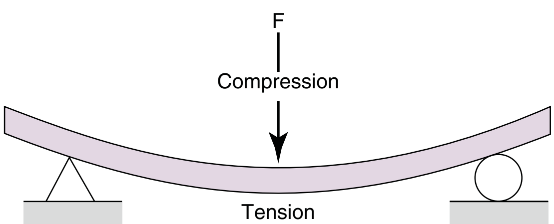


Figure 2.10 Flexure. *F*, Force.

Yield Stress

The yield stress or yield point is the point at which the material begins to maintain a permanent deformational change. This change is known as "plastic" deformation. At stresses under the yield point, a material will always return to its original shape once the load is removed. This is described as the elastic range of the material.

Ultimate Stress

Ultimate stress is the stress at which a material ruptures. The strength of the material before it ruptures also depends on the type of stress to which it is subjected. For example, ultimate shear stresses are usually lower than ultimate tensile stresses (i.e., less shear stress must be applied before the material ruptures than in the case of tensile or compressive stress).

Strain

Materials subjected to any stress will become deformed or change in shape, even at very small levels of stress. If a material lengthens or shortens in response to stress, it is said to experience *strain*. Strain is denoted by ε and can be found by dividing the total elongation (or contraction) ΔL by the original length L_O of the structure being loaded:

$$\varepsilon = \frac{\Delta L}{-L_0} \quad (2.2)$$

Consider a change in length ΔL of a wire or rod caused by a change in stretching force F (Fig. 2.11). The amount of stretch is proportional to the original length of wire.

Stress-Strain Curve

The most widely used means of determining the mechanical properties of materials is the tension test. Much can be learned from observing the data collected from such a test. In the tension test, the shape (dimensions) of the test specimen are fixed by standardization so that the results can be universally understood, no matter where or by whom the test is conducted. The test specimen is mounted between the jaws of a tensile testing machine, which is simply a device used to stretch the specimen at a controlled rate. As defined by standards, the cross-sectional area of the test specimen is smaller in the center to prevent failures where the test specimen is gripped. The specimen's resistance to being stretched and the linear deformations are measured using sensitive instrumentation (Fig. 2.12).

The force of resistance divided by the cross-sectional area of the specimen is the *stress* in the specimen (Eq. 2.1). The *strain* is the total deformation divided by the original length (Eq. 2.2). The stress and the

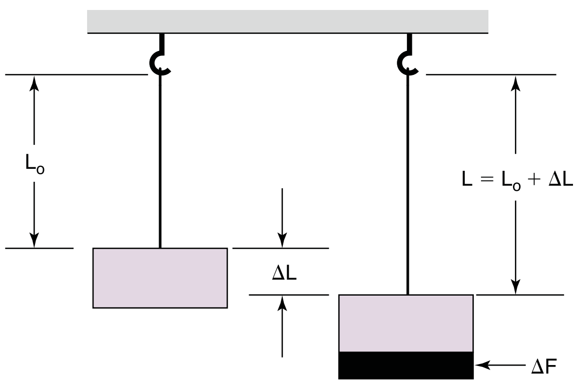


Figure 2.11 Strain. *F*, Force; *L*, length.

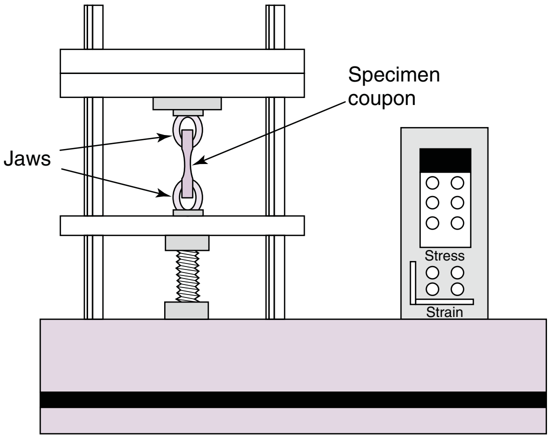


Figure 2.12 Tension test.

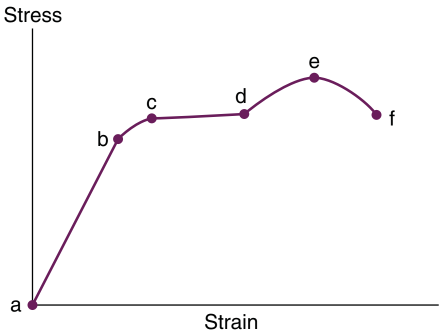


Figure 2.13 Stress-strain.

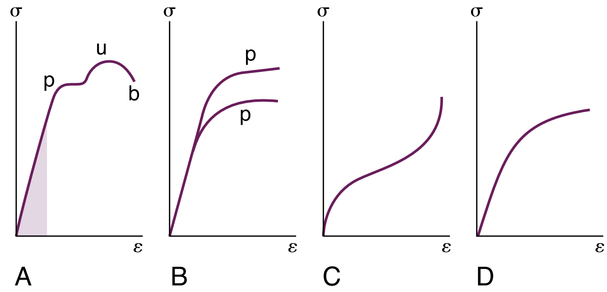


Figure 2.14 (A–D) Stress (σ) - Strain (ε) diagrams for different materials.

corresponding strain of a material undergoing a tensile test can be shown on a stress strain curve, with stress represented on the Y axis and strain represented on the X axis. Fig. 2.13 shows a typical stress-strain diagram for a mild steel specimen.

The shape and magnitude of the stress-strain curve of a metal depend on its composition; heat treatment; history of plastic deformation; and strain rate, temperature, and state of stress imposed during testing. The parameters used to describe the stress-strain curve of a metal are tensile strength, yield strength or yield point, percent elongation, and reduction in area. The first two are strength parameters; the last two indicate ductility, or the material's ability to be stretched (and remain stretched) under tension.

The general shape of the stress-strain curve (Fig. 2.13) requires further explanation. In the region from *a* to *b*, the stress is linearly proportional to the strain, and the strain is elastic (i.e., the stressed part returns to its original shape when the load is removed). When the applied stress exceeds the yield strength, *b*, the specimen undergoes plastic deformation. If the load is subsequently reduced to zero, the part remains permanently deformed. The stress required to produce continued plastic deformation increases with increasing plastic strain (points *c, d,* and *e* in Fig. 2.13)—that is, the metal strain hardens. The volume of the part remains constant during plastic deformation, and as the part elongates, its cross-sectional area decreases uniformly along its length until point *e* is reached. The ordinate of point *e* is the tensile strength of the material. After point *e,* further elongation or deformation occurs with proportionally lower amounts of stress until the part ruptures at point *f* (breaking or fracture strength).

Stress-strain diagrams assume widely differing forms for various materials. Fig. 2.14A shows the stress-strain diagram for a medium-carbon structural steel. The ordinates of points *p, u,* and *b* are the yield point, tensile strength, and breaking strength, respectively. The lower curve of Fig. 2.14B is for an alloy steel, and the higher curve is for hard steels. Nonferrous alloys and cast iron have the form shown in Fig. 2.14C. The plot shown in Fig. 2.14D is typical for rubber. Note that these are representative graphs only. The dimensions (and scale) vary greatly for the materials mentioned here.

For any material with a stress-strain curve of the form shown in Fig. 2.14, it is evident that the relationship between stress and strain is linear for comparatively small values of the strain. This linear relationship between elongation and the axial force causing it was first reported by Sir Robert Hooke in 1678 and is called *Hooke's law.* Expressed as an equation, Hooke's law becomes:

$$\sigma = \varepsilon_E \qquad (2.3)$$

where σ = stress (psi), ε = strain (inch/inch), and *E* = constant of proportionality between stress and strain. This constant is also called *Young's modulus* or the *modulus of elasticity.*

The slope of the stress-strain curve from the origin to point *p* (Fig. 2.14A and B) is the modulus of elasticity of that particular material *E.* The region where the slope is a straight line is called the *elastic region,* where the material behaves in what we typically consider an elastic manner; that is, it is loaded and stretched, and upon releasing the load, the material returns to its original position. The ordinate of a point coincident with *p* is known as the *elastic limit* (i.e., the maximum stress that may develop during a simple tension test such that no permanent or residual deformation occurs when the load is entirely removed). Values for *E* are given in Table 2.1. Devices and materials are designed to perform in the elastic region (with very few exceptions).

In a routine tension test (Fig. 2.15), which illustrates Hooke's law, a bar of area *A* is placed between two jaws of a vise, and a force *F* is applied to compress the bar. Combining Eqs. 2.1, 2.2, and 2.3 and solving for the shortening Δ*L* gives:

$$\Delta L = \frac{FL_0}{AE} \qquad (2.4)$$

Because the original length L_O, cross-sectional area *A*, and modulus of elasticity *E* are constants, the shortening Δ*L* depends solely on *F.* As *F* doubles, so does Δ*L.*

The operation of a steel spring scale is another practical illustration of Hooke's law (Fig. 2.16). The amount of deflection of the spring for every unit of force of the load remains constant. In Fig. 2.16A, the scale indicates three units (pounds, ounces, or grams). With one weight added (Fig. 2.16B), the scale indicates 5, or two additional units. A second weight added (Fig. 2.16C) causes the scale to indicate 7, or a total of four additional units, and a third weight stretches the spring two more units (Fig. 2.16D). Therefore it is possible to make uniform

TABLE 2.1 Modulus of Elasticity

Material	E(×10⁶ psi)	E (GPa)	Material	E(×10⁶ psi)	E (GPa)
Steel	30	200	Magnesium	6.5	45
Carbon composite	18.5	130	Bone	2.85	20
Copper	16	110	Polyester-Dacron	2	14
Brass	15	105	Polyester (resin)	0.65	4.5
Bronze	12	85	Surlyn (ionomer)	0.34	2.5
Aluminum	10.3	70	Polypropylene	0.23	1.6
Kevlar	9	62	High-density polypropylene	0.113	0.8
Glass	8.4	58	High-density polypropylene	0.018	0.13

GPa, Gigapascals; *psi,* pounds per square inch.

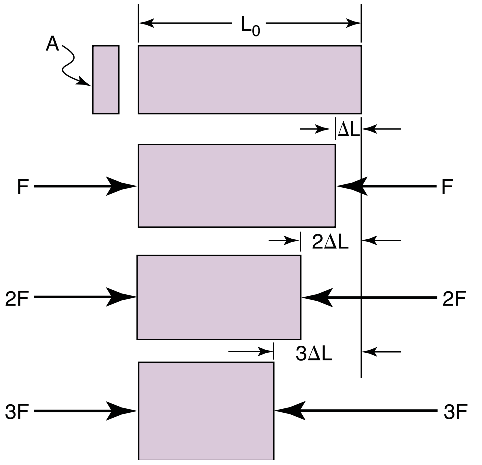


Figure 2.15 Linearity. *A,* cross-sectional area; *F,* force; *L,* length.

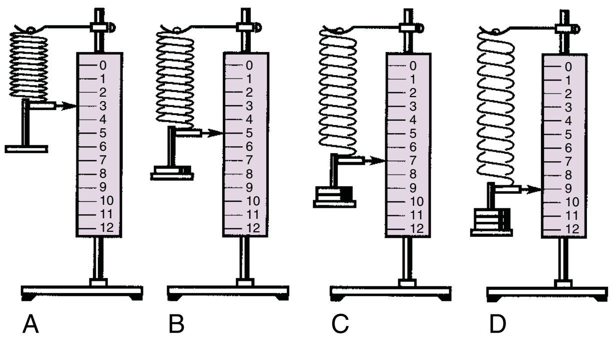


Figure 2.16 (A–D) Linear relationship between stretch and weight.

gradations for every unit of force to the point beyond the range of elasticity where the spring would distort or break. Scales are manufactured with springs strong enough to bear predetermined maximum loads. A compression spring scale designed to remain within the elastic range, recording weights to about 250 lb (100 kg) and then returning back to 0, is the type commonly used for weighing people.

Plastic Range

Plastic range is beyond the elastic range (*b* to past *e* on the stress-strain diagram of Fig. 2.13), and the material behaves plastically. That is, the material has a set or permanent deformation when externally applied loads are removed; it has "flowed" or become plastic. In the case of the steel spring scale, if the weight did not actually break the spring, it

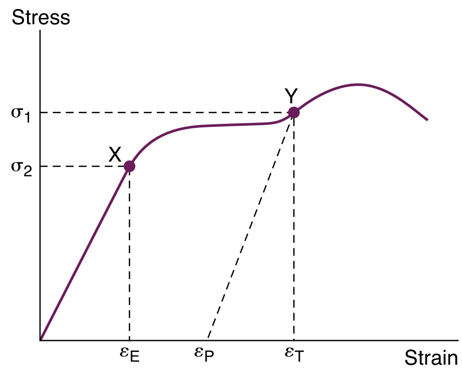


Figure 2.17 Plastic strain.

would stretch it permanently so that the readings on the scale would no longer be accurate.

For most materials, the stress-strain curve has an initial linear elastic region in which deformation is reversible. Note the load σ_2 in Fig. 2.17. This load will cause strain ε_E. When the load is removed, the strain disappears; that is, point *X* (σ_2, ε_E) moves linearly down the proportional part of the curve to the origin. Similarly, when load σ_1 is applied, strain ε_T results. However, when load σ_1 is removed, point *Y* does not move back along the original curve to the origin but moves to the strain axis along a line parallel to the original linear region intersecting the strain axis at ε_P. Therefore with no load, the material has a residual or permanent strain of ε_P. Plastic deformation is difficult to judge but can be predicted for sidebars and charted as previously mentioned. The quantity of permanent strain ε_P is the plastic strain, and ($\varepsilon_T - \varepsilon_P$) is the elastic strain ε_E or:

$$\varepsilon_T - \varepsilon_P = \varepsilon_E \quad (2.5)$$

where ε_T = total strain under load, ε_P = plastic (or permanent) strain, and ε_E = elastic strain.

Yield Point

Yield point (point *b* on the stress-strain diagram of Fig. 2.13) refers to that point at which a marked increase in strain occurs without a corresponding increase in stress. The horizontal portion of the stress-strain curve (*b-c-d* in Fig. 2.13) indicates the yield stress corresponding to this yield point. The yield point is the "knee" in the stress-strain curve for a material and separates the elastic from the plastic portions of the curve. In most all material applications, efforts are made to optimize the cross-sectional area of the material to ensure that the maximum stress applied during use of the orthosis or prosthesis falls below the yield point and comfortably within the elastic range of the material.

This ensures that the orthosis or prosthesis, once worn by the patient, maintains a consistent shape. This is true even if some plastic deformation was required to initially shape the material to the model.

Tensile Strength

The tensile strength of a material is obtained by dividing the maximum tensile force reached during the test (*e* on the stress-strain diagram in Fig. 2.13) by the original cross-sectional area of the test specimen. Practical application of the maximal tensile force is minimal because devices are never designed to be loaded to this value.

Toughness and Ductility

The area under the curve to the point of maximum stress (*a-b-c-d-e* in Fig. 2.13) indicates the *toughness* of the material, or its ability to withstand shock loads before rupturing. The supporting arms of a car bumper are an example of where toughness is of great value as a mechanical property. *Ductility,* as stated earlier, is the ability of a material to sustain large permanent deformations in tension (i.e., to be stretched), such as when drawing a rod into a wire. The distinction between ductility and toughness is that ductility deals only with the ability to deform, whereas toughness considers both the ability to deform and the stress developed during the deformation. The requirement for plastic deformation in sidebars is weighed against the ability of the sidebars to resist large rapid loads and even the forces required by the practitioner to deform them.

Thermal Stress

When a material is subjected to a change in temperature, its dimensions increase or decrease as the temperature rises or falls. If the material is constrained by neighboring structures, stress is produced.

The influence of temperature change is noted by a term called the coefficient of thermal expansion α, which is defined as the unit of strain produced by a temperature change of 1 degree. This physical constant is a mechanical property of each material. Values of α for several materials are given in Table 2.2.

If the temperature of a bar of length L_O inches is increased ΔT F (or C; *NOTE:* α indicates which measure of temperature it relates to), the elongation ΔL in any units of the unrestrained bar is given by:

$$\Delta L = \alpha L_0 \Delta T \quad \textbf{(2.6)}$$

If the heated rod is compressed back to its original length, it will experience compression as given by Eq. 2.4:

$$\Delta L = \frac{FL_0}{AE} \quad \textbf{(2.7)}$$

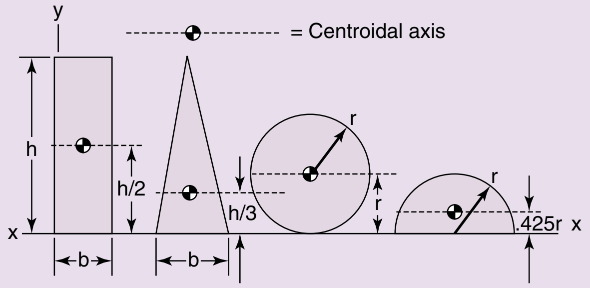
Combining Eqs. 2.6 and 2.7 and solving for stress, $\sigma = F/A$, gives:

$$\sigma = \alpha \Delta TE \quad \textbf{(2.8)}$$

Eq. 2.8 allows the calculation of stress in a rod as a function of the increase in temperature ΔT, the modulus of elasticity E (Table 2.1), and the coefficient of thermal expansion α (Table 2.3).

TABLE 2.2 Geometric Factors for Common Shapes

	Rectangle	Triangle	Circle	Semicircle
$\bar{y}_c$	$h/2$	$h/3$	r	$0.425r$
I_{cc}	$bh^3/12$	$bh^3/36$	$0.785r^4$	$0.11r^4$
I_{xx}	$bh^3/3$	$bh^3/12$	$3.93r^4$	$0.393r^4$
Z	$bh^2/6$	$bh^3/24$	$0.785r^3$	$0.19r^3$
C	$h/2$	$2h/3$ (top) $h/3$ (bottom)	r	$0.575r$ (top) $0.425r$ (bottom)



Centroids and Center of Gravity

The centroid and center of gravity of objects play important roles in the objects' mechanical properties. The center of gravity and centroid of two identically shaped objects are the same if the density is uniform in each object. The centroid is a geometric factor, and the center of gravity depends on mass.

For an object of uniform density, the term *center of gravity* is replaced by the *centroid of the area.* The *centroid of an area* is defined as the point of application of the result of a uniformly distributed force acting on the area. An irregularly shaped plate of material of uniform thickness *t* is shown in Fig. 2.18. Two elemental areas (*a* and *b*) are shown with centroids (x_1,y_1) and (x_2,y_2), respectively. If the large, irregularly shaped plate is divided into small elemental areas, each having its own centroid, then the centroid for the irregularly shaped plate is (x,y), where:

$$\bar{x} = \frac{\bar{x}_i \bar{a}_i}{\Sigma_i A}$$

$$\bar{y} = \frac{\bar{y}_i \bar{a}_i}{\Sigma_i A}$$

and

$$\bar{x} = \frac{\bar{x}_1 \bar{a}_1 + \bar{x}_2 \bar{a}_2 + \cdots}{\Sigma_i A}$$

$$\bar{y} = \frac{\bar{y}_1 \bar{a}_1 + \bar{y}_2 \bar{a}_2 + \cdots}{A}$$

TABLE 2.3 Coefficient of Thermal Expansion

Material	Coefficient α ($\times 10^{-6}$ per °F)	Coefficient α ($\times 10^{-6}$ per °C)	Material	Coefficient α ($\times 10^{-6}$ per °F)	Coefficient α ($\times 10^{-6}$ per °C)
Steel	6.5	11.7	Brass	10.4	18.7
Cast iron	6	10.8	Bronze	10	18
Wrought iron	6.7	12	Aluminum	12.5	22.5
Copper	9.3	16.7	Magnesium	14.5	26.1

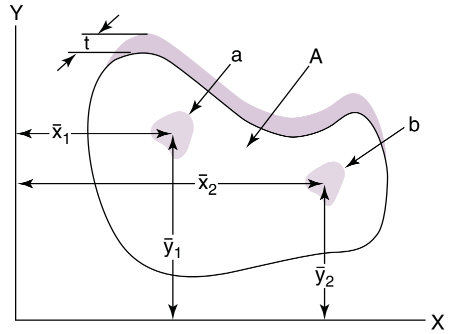


Figure 2.18 Centroids. *A,* The total area of the object; *a,* any small area which is part of A; *b,* any other small area that is part of A. *t,* thickness.

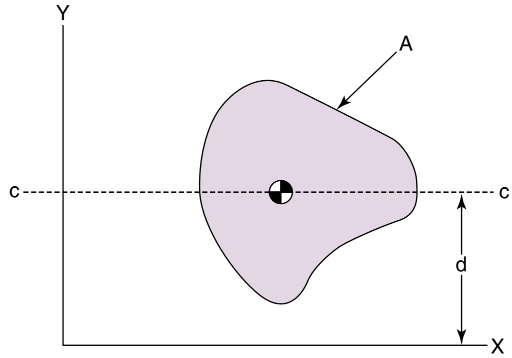


Figure 2.19 Parallel axis theorem. *A,* The area of the object; *c;* an axis that passes through the centroid and is parallel to the X axis at some distance; *d,* away from the X axis.

The *y*-centroids for several common geometric shapes are given in Table 2.2.

MOMENT OF INERTIA

The moment of inertia of a finite area about an axis in the plane of the area is given by the summation of the moments of inertia about the same axis of all elements of the area contained in the finite area. In general, the *moment of inertia* is defined as the product of the area and the square of the distance between the area and the given axis. The moments of inertia about the centroidal axes I_{cc} of a few simple but important geometric shapes are determined by integral calculus and are given in Table 2.2. Although Young's modulus is an indication of the strength of the material, the moment of inertia is an indicator of the strength of a particular shape about a particular axis. A shape will have a different moment of inertia depending on how the load is applied. An example of this is a long, thin rectangle. The rectangle is "weaker," or easier to bend, if bent along its length; however, it is "stronger" if it is bent about its height. This is a highly important parameter for the practitioner to know, as the shape of an object can be altered far more than the strength of the materials being used.

Parallel Axis Theorem

When the moment of inertia has been determined with respect to a given axis, such as the centroidal axis, the moment of inertia with respect to a parallel axis can be obtained by the *parallel axis theorem,* provided one of the axes passes through the centroid of the area. The parallel axis theorem states that the moment of inertia with respect to any axis is equal to the moment of inertia with respect to a parallel axis through the centroid added to the product of the area and the square of the distance between the two axes (Fig. 2.19):

$$I_{xx} = I_{cc} + Ad^2$$

or

$$I_{cc} = I_{xx} - Ad^2 \qquad \textbf{(2.9)}$$

where I_{xx} = moment of inertia about the *x*-axis, I_{cc} = moment of inertia about the centroid, A = area, and d = distance between axes.

Stresses in Beams

If forces are applied to a beam, as shown in Fig. 2.20, downward bending of the beam occurs. It is helpful to imagine a beam is composed of an infinite number of thin longitudinal rods or fibers. Each longitudinal fiber is assumed to act independently of every other fiber (i.e., there are no lateral stresses [shear] between fibers). The beam of Fig. 2.20 will deflect downward, and the fibers in the lower part of the beam will undergo extension, whereas those in the upper part will shorten. Changes in the lengths of the fibers set up stresses in the fibers. Those that are extended have tensile stresses acting on the fibers in the direction of the longitudinal axis of the beam, whereas those that are shortened are subject to compression stresses.

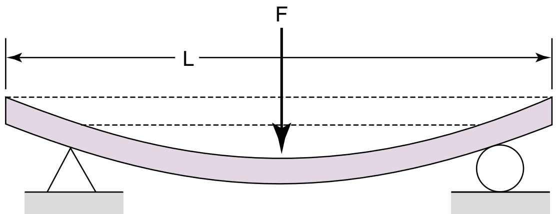


Figure 2.20 Beam stress. *F,* Force; *L,* length.

One surface in the beams always contains fibers that do not undergo any extension or compression and are thus not subject to any tensile or compressive stress. This surface is called the *neutral surface* of the beam. The intersection of the neutral surface with any cross-section of the beam perpendicular to its longitudinal axis is called the *neutral axis.* All fibers on one side of the neutral axis are in a state of tension, whereas those on the opposite side are in compression.

For any beam having a longitudinal plane of symmetry and subject to a bending torque T at a certain cross-section, the normal stress σ, acting on a longitudinal fiber at a distance y from the neutral axis of the beam (Fig. 2.21), is given by:

$$\sigma = \frac{Ty}{I} \qquad \textbf{(2.10)}$$

where I = moment of inertia of the cross-sectional area about the neutral or centroidal axis in in^4, or (m^4).

These stresses vary from zero at the neutral axis of the beam ($y = 0$) to a maximum at the outer fibers (Fig. 2.21). These stresses are called *bending, flexure,* or *fiber stresses.*

Section Modulus

The value of y at the outer fibers of the beam is typically denoted by c. At these fibers, the bending stress is at a maximum and is given by:

$$\sigma = \frac{Tc}{I} = \frac{Tc}{I/c} \qquad \textbf{(2.11)}$$