JHT READ FOR CREDIT ARTICLE #217.

Understanding the Biomechanical Nature of Musculoskeletal Tissue

GUEST EDITOR COMMENT: *Although hand therapists infrequently use terms such as stress, strain, force, moments, or Young's modulus, these concepts are used every day in a hand therapy practice. Such concepts are the underlying reason why we incorporate treatment techniques such as orthotic devices, exercises, and the use of adaptive equipment into hand therapy protocols. To gain a deeper understanding of the basic science underlying our treatment technique, a working knowledge of such terms and the biomechanical nature of musculoskeletal tissue is necessary. This introductory article lays the foundation of these concepts. These concepts, as they relate to specific tissues and treatment techniques, are discussed throughout this Special Issue.*—VICTORIA PRIGANC, PhD, OTR, CHT, CLT, *Guest Editor*

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ABSTRACT: This article provides a general overview of the biomechanical principles associated with hand therapy. Specifically, it reviews the basic topics of material properties (including both theoretical principles and practical concepts), static analysis (including forces, moments, muscle forces, and Newton's laws), and ends with a clinical example involving analysis of the risk of damage to the A3 pulley.

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MATERIAL PROPERTIES

Theoretical Concepts for Material Properties

Stress (σ) is defined as the force (F) acting on an object, divided by the cross-sectional area (A) over which it is acting:

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

The units of stress are in Pascals (N/m^2) in the metric system and psi (pounds per square inch) in the English system. In general, there are three main forms of stress to be considered for biological tissues: tensile, compressive, and shear stresses (Figure 1). More complex loading conditions, such as bending, can result in a combination of tensile and compressive stresses. A simple example may help demonstrate the calculation of stress. Consider a situation in which a 100-N compression force is being exerted along the long axis of a bone, which has an outer diameter of 10 cm and an inner diameter of 6 cm (Figure 2). The stress would be equal to the force (100 N) divided by the cross-sectional area (0.02 m^2), or 5000 Pascals.

Strain (ε) can be defined as the change in length (ΔL) of an object divided by the original length (Lo):

$$\varepsilon = \frac{\Delta L}{\text{Lo}}$$

Strain is typically reported as having no units or in percentage. Very small strains are reported in microstrains, where 1 million microstrains would represent a strain level of 1. As an example, if a tendon starts with a length of 5 cm and then is stretched to a new length of 5.1 cm, that would represent a strain of 0.02 or 2%.

A good way to consider stress and strain is that they represent a model for normalizing the loads and deformations on a body based on its size. Consider a rectangular object with a given length, height, and width. To determine the stress acting on this object, the force is normalized by dividing by the height × width. To determine the strain, the deformation is normalized by the length. Thus, all three dimensions of the object are taken into account.

We will next consider the relationship between the loads applied to a body and the resultant deformations. In general, as loads are applied, the deformation will continue to increase. This could result in a decrease in length (compression) or an increase in length (tension), depending on the direction of loading. For many materials, there is a linear relationship

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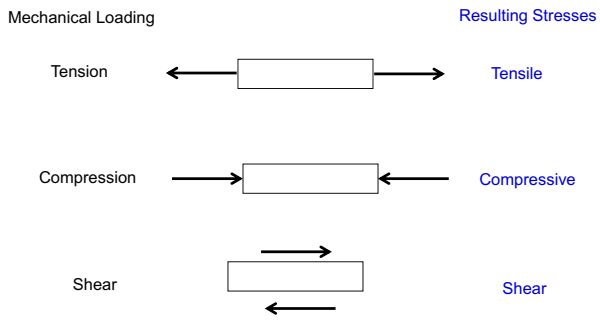


FIGURE 1. Mechanical loading and resultant stresses for the three common loading conditions for biological materials.

between the applied force and resulting deformation and the slope of this curve represents the stiffness of the material (Figure 3). A material that is harder to deform for a given level of force is considered a stiffer material. This linear relationship occurs in the elastic region, where the tissue is not damaged and returns to its original length after the load is removed. If the load is large enough, the tissue will pass a certain yield point and enter the plastic region, indicating that there is some tissue damage. As the load is continually increased it will eventually reach the ultimate yield point, which is the highest load that can be borne by the tissue before it breaks in two. The relationship between these terms is illustrated in Figure 4.

If we consider three structures, all composed of similar material, but with three different cross-sectional areas, stiffness will increase with increasing area (Figure 5A). However, if we normalize the force—

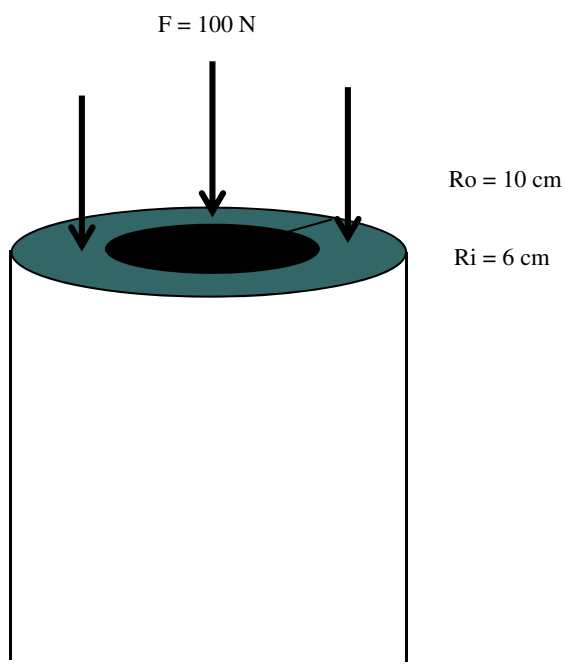


FIGURE 2. Distributed load along the long axis of a bone—see text for analysis.

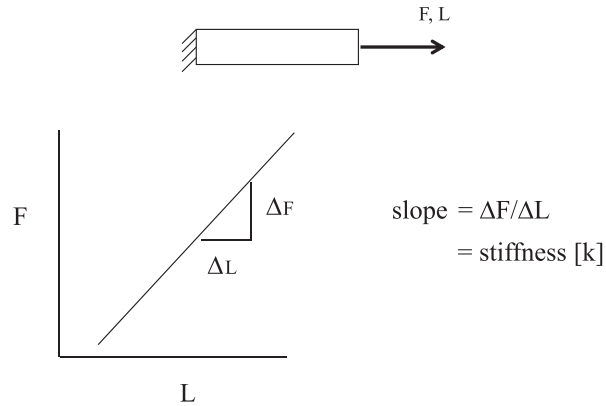


FIGURE 3. The slope of a linear force—length relationship represents the stiffness of a material.

length relationship and turn it into a stress—strain relationship, then the three curves will sit on top of each other (Figure 5B). The slope of the stress—strain curve is referred to as the modulus of elasticity, or Young's modulus. It is important to point out the fundamental difference between stiffness and Young's modulus. Stiffness is considered a structural property and is dependent on both material composition and size. However, Young's modulus is a material property, so it is only dependent on the material composition. Another way to look at this is to consider the following two items: Aluminum foil and an aluminum baseball bat. Clearly, the aluminum foils is far less stiff, as it can be deformed with very little force. However, if one were to take very accurate force—length measurements and convert those to stress—strain measurements, these curves would be very similar, with a resulting similar Young's modulus for the foil and bat.

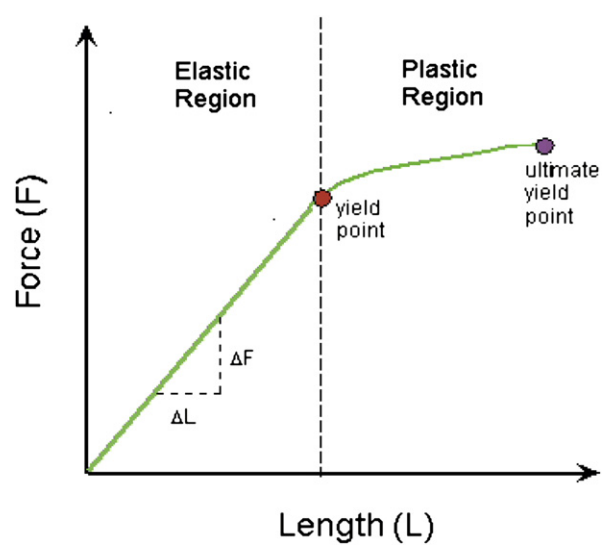


FIGURE 4. Graph demonstrating a linear force—length relationship in the elastic region. Once the yield point is reached, the material is being damaged and enters the plastic region. Once the ultimate yield point is reached, the material ruptures and can no longer bear any load.

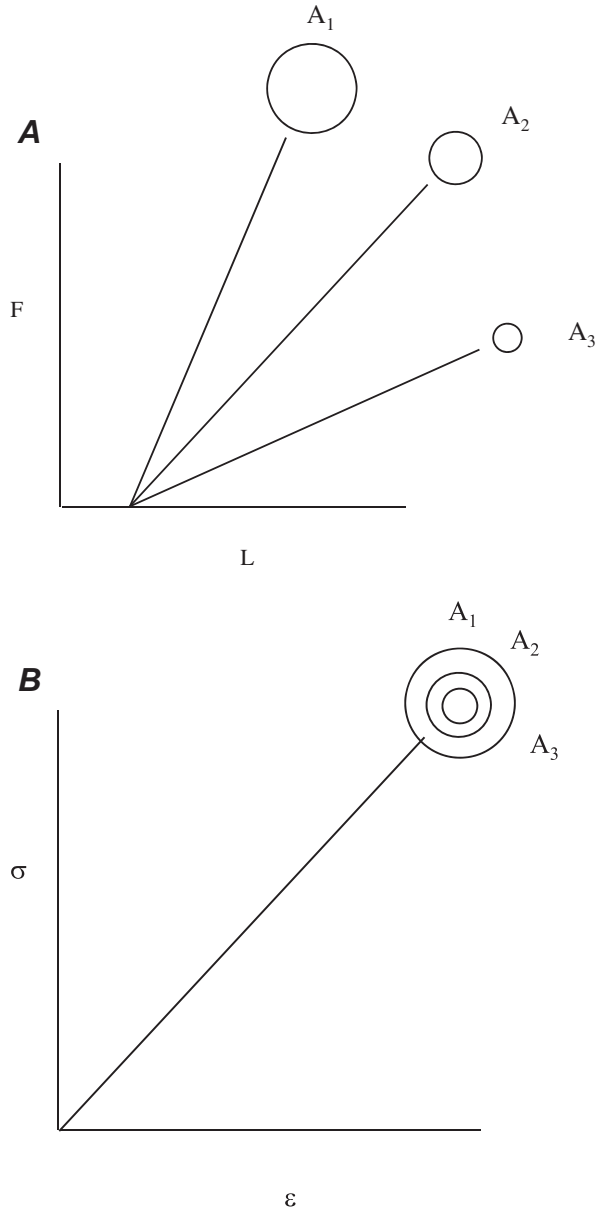


FIGURE 5. Loading response of three structures composed of the same material, but with differing cross-sectional areas. ***(A)*** *Stiffness, or the slope of the force—length curve, increase with increasing cross-sectional area.* ***(B)*** *Young's modulus, or the slope of the stress—strain curve, is independent of cross-sectional area.*

Material Properties of Biological Tissues

The material properties of tendons and ligament are almost entirely dictated by the wavy collagen fibers that run throughout these tissues, which is parallel in tendons and more irregular in ligaments. It is generally accepted that these properties are a result of the straightening out (or uncrimping) of these fibers. As the strain in the tissue is gradually increased, more and more collagen fibers are recruited. This results in a nonlinear stress—strain relationship, with an increase in the modulus of elasticity at higher strain levels.

As with tendons and ligaments, the material properties of articular cartilage are largely determined by collagen. With cartilage however, it is also important to consider that it is embedded in proteoglycans, which are hydrophilic. So when the tissue is not loaded, it tends to swell with fluid. As it is loaded, the proteoglycans compress, which results in an increase in hydrostatic pressure within the tissue, which in turn results in tensile loading of the cartilage and outflow of fluid into the synovial fluid.

Like most connective tissue, bone contains collagen fibers that are embedded in proteoglycans. However, to a large extent, the constitutive properties of bone are controlled by the inorganic mineral component, consisting mainly of hydroxyapatite, which is a combination of calcium and phosphate. This is what contributes to its linear stress—strain relationship and high Young's modulus. There are two main types of bone, with very distinct material properties. Cortical (or compact) bone surrounds the outside of long bones, whereas cancellous (or trabecular bone) is contained in the inside of the bone. Cortical bone tends to have a much high bone density, which results in a higher modulus of elasticity. The distinction between these two types of bone can be easily viewed on an X-ray of the hand, where the outer regions of the bones are a much brighter shade of white (Figure 6).

STATICS

The musculoskeletal system is responsible for generating forces that move the human body in space and prevent unwanted motion. Understanding the mechanics and pathomechanics of human motion requires an ability to study the forces and moments applied to, and generated by, the body or a particular body segment.

Forces

For the purpose of the musculoskeletal system, a force can be defined as a "push or pull" that results from physical contact between two objects. The only important exception to this rule is the force due to gravity, in which there is no direct physical contact between the two objects. Some of the more common force generators with respect to the musculoskeletal system include muscles/tendons, ligaments, friction, ground reaction, and weight. A force is a vector quantity with magnitude, orientation, direction, and a point of application. In general, the point of application of a force (e.g., tendon insertion) is located with respect to a fixed point on a body, usually the joint center of rotation. This information is used to calculate the moment due to that force.

FIGURE 6. *Typical X-ray of a human hand (provided by Jason Tavakolian, MD, Slocum Center for Orthopedics and Sports Medicine).*

Moments

A moment is typically caused by a force acting at a distance from the center of rotation of a segment. A moment tends to cause a rotation and is defined by the cross product function: $M = r \times F$. Therefore, a moment is represented by a vector that passes through the point of interest (e.g., the center of rotation) and is perpendicular to both the force and distance vectors. For a two-dimensional analysis, both the force and the distance vectors are in the plane of the paper, so the moment vector is always directed perpendicular to the page, with a line of action through the point of interest. Because it has only this one orientation and line of action, a moment is often treated as a scalar quantity in a two-dimensional analysis, with only magnitude and direction. Torque is another term that is synonymous with a scalar moment. From the definition of a cross product, the magnitude of a moment (or torque) is calculated as $M = r \cdot F \cdot \sin(\theta)$. Its direction is referred to as the direction in which it would tend to cause an object to rotate. Although there are several different distances that can be used to connect a vector and a point, the same moment is calculated no matter which distance is selected. The distance that is perpendicular to the force vector is referred to as the moment arm (MA) of that force. Because the sine of 90° is equal to 1, the use of a MA simplifies the calculation of moment to $M = MA \times F$. The moment can also be calculated from any distance as $MA \times r \times \sin(\theta)$. In general, muscles are responsible for producing both forces and moments, thus resulting in both translational and rotational motion.

Muscle Forces

As mentioned previously, there are three important parameters to consider with respect to the force of a muscle: orientation, magnitude, and point of application. With some care, it is possible to measure orientation and line of action from cadavers or imaging techniques such as magnetic resonance imaging and computed tomography. This information is helpful in determining the function and efficiency of a muscle in producing a moment. These analyses are useful because they can be performed even if the magnitude of a muscle's force is unknown. However, to understand a muscle's function completely, its force magnitude must be known. Although forces can be measured with invasive force transducers, instrumented arthroplasty systems, or simulations in cadaver models, there are currently no noninvasive experimental methods that can be used to measure the in vivo force of intact muscles. Consequently, basic concepts borrowed from freshman physics can be used to predict muscle forces. Although they often involve many simplifying assumptions, such methods can be very useful in understanding joint mechanics.

Newton's Laws

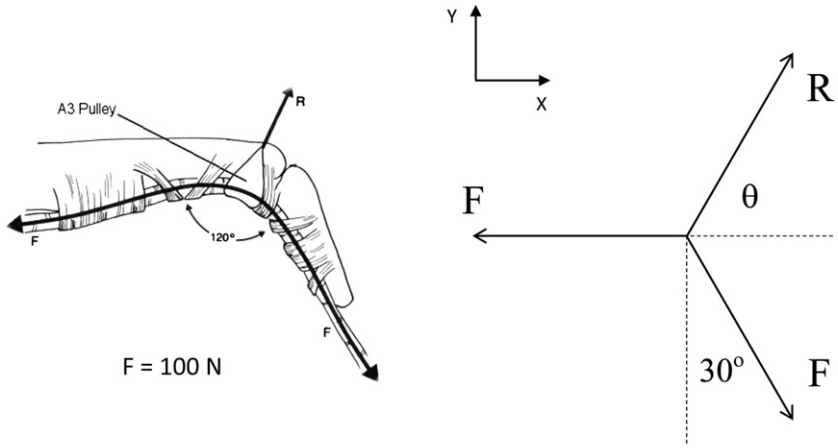
Statics is the study of the forces acting on a body at rest or moving with a constant velocity. Although the human body is almost always accelerating, a static analysis offers a simple method of addressing musculoskeletal problems. This analysis may either solve the problem or provide a basis for a more sophisticated dynamic analysis. Because the musculoskeletal system is simply a series of objects in contact with each other, some of the basic physics principles developed by Sir Isaac Newton are useful. Newton's laws are as follows:

First law: An object remains at rest (or continues moving at a constant velocity) unless acted upon by an unbalanced external force.

Second law: If there is an unbalanced force acting on an object, it produces an acceleration in the direction of the force, directly proportional to the force ($f = ma$).

Third law: For every action (force), there is a reaction (opposing force) of equal magnitude but in the opposite direction.

From Newton's first law, it is clear that if a body is at rest, there can be no unbalanced external forces acting on it. In this situation, termed static equilibrium, all of the external forces acting on a body must add (in a vector sense) to zero. An extension of this law



Sum forces in x and y directions:

$\Sigma F_X = 0$	$\Sigma F_Y = 0$
$R_X - F + F \sin(30^\circ) = 0$	$R_Y - F \cos(30^\circ) = 0$
$R_X = F - F \sin(30^\circ)$	$R_Y = F\cos(30^\circ)$
$R_X = 100 - 100 \sin(30^\circ)$	$R_Y = 100 \cos(30^\circ)$
$R_X = 50$ N	$R_Y = 87$ N

Resultant Force:

$$R = \sqrt{(R_X)^2 + (R_Y)^2} = \sqrt{50^2 + 87^2} = 100\ N$$

$$\theta = \tan^{-1}\left(\frac{R_Y}{R_X}\right) = \tan^{-1}\left(\frac{87}{50}\right) = 60°$$

FIGURE 7. *Biomechanical analysis of the load acting on the A3 pulley during a theoretical pinching activity. (Adapted with permission. Musculoskeletal System. 3rd ed. Lippincott Williams and Wilkins, 2001.)*

to objects larger than a particle is that the sum of the external moments acting on that body must also be equal to zero for the body to be at rest. Therefore, for a three-dimensional analysis, there are a total of six equations that must be satisfied for static equilibrium:

$$\Sigma F_x = 0 \quad \Sigma F_y = 0 \quad \Sigma F_z = 0$$

$$\Sigma M_x = 0 \quad \Sigma M_y = 0 \quad \Sigma M_z = 0$$

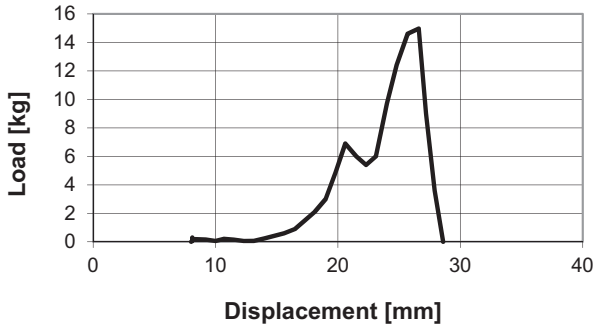
For a two-dimensional analysis (in the X-Y plane), there are only two in-plane force components and one perpendicular moment (torque) component:

$$\Sigma F_x = 0 \quad \Sigma F_y = 0 \quad \Sigma M_z = 0$$

Under many conditions, it is reasonable to assume that all body parts are in a state of static equilibrium and these three equations can be used to calculate some of the forces acting on the musculoskeletal system. When a body is not in static equilibrium, Newton's second law states that any unbalanced forces and moments are proportional to the acceleration of the body. That situation is beyond the scope of this article.

Clinical Example

Therapists can use basic biomechanics information to estimate forces and the result of such forces on tissues. For example, published information exists regarding forces exerted on tendons when performing different activities. Using this published information along with knowledge of biomechanical principles and disease processes, the therapist can estimate the effects of different activities, exercises, or splinting forces on tissues. For example, under normal conditions, the A3 pulley of the finger provides for a smooth turn of the long flexor tendons at the proximal interphalangeal joint. However, in patients with rheumatoid arthritis, this structure may become



$F_{YIELD} = (7\ kg) * (10\ m/s^2) = 70\ N$

$F_{ULT\ YIELD} = (15\ kg) * (10\ m/s^2) = 150\ N$

FIGURE 8. *Simulated force—length curve of the A3 pulley.*

weakened. For this problem, consider a simple pinching activity, where the force (*F*) acting on the flexor tendons is estimated to be 100 Newtons, as shown in Figure 7. Given that information, the distraction force (*R*) acting on the A3 pulley can be calculated as shown. To determine the significance of this load, consider the load—displacement curve from an A3 pulley on a human cadaver that had rheumatoid arthritis (Figure 8). Based on these data and analysis, it appears that the pulley is in danger of being damaged, but not ruptured.

Suggested Readings in Biomechanics

Hamill J, Knutzen KM. Biomechanical Basis of Human Movement. 3rd ed. Philadelphia: Lippincott Williams and Wilkins, 2009.

Hall SJ. Basic Biomechanics. 6th ed. Boston: McGraw Hill, 2012.

Nordin M, Frankel VH. Basic Biomechanics of the Musculoskeletal System. 3rd ed. Philadelphia: Lippincott Williams and Wilkins, 2001.

WhitingW, Zernicke R. Biomechanics of Musculoskeletal Injury. 2nd ed. Champaign: Human Kinetics, 2008.

McGinnis P. Biomechanics of Sport and Exercise. 2nd ed. Champaign: Human Kinetics, 2004.

Oatis CA. Kinesiology: The Mechanics and Pathomechanics of Human Movement. Baltimore: Lippincott Williams and Wilkins, 2009.

Neumann DA. Kinesiology of the Musculoskeletal System: Foundations for Physical Rehabilitation. 2nd ed. St. Louis: Mosby, 2009.

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Quiz: Article #217

Record your answers on the Return Answer Form found on the tear-out coupon at the back of this issue or to complete online and use a credit card, go to *JHTReadforCredit.com*. There is only one best answer for each question.

#1. The three main forms of stress applied to biological tissues are
   a. torque, strain, and viscoelastic
   b. tensile, compressive, and shear
   c. elongating, shrinking, and expanding
   d. friction, heat, and rupture

#2. Strain is defined as
   a. change in length
   b. change in length divided by unit of torque
   c. change in length times unit of torque
   d. change in length divided by original length

#3. The ultimate yield point is the
   a. active end range
   b. passive end range
   c. maximum load a tissue can bear before breaking
   d. maximum amount of strain a tissue can bear before pain stops the load

#4. Elongation of collagen tissue (e.g. tendon, ligament) is attributed to the phenomenon of
   a. stress elongation
   b. uncrimping
   c. hysteresis
   d. preconditioning

#5. The author believes that hand therapists seldom use terms such as stress, strain, force, or moments, yet use those same concepts routinely in the clinic
   a. true
   b. false

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