



Fundamentals of biomechanics

Educational article - English edition

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Source article: Basic Biomechanics. David J. Giurintano, Journal of Hand Therapy, 1995.

This educational article presents the biomechanical principles needed to analyse movement, design orthoses and dose tissue loading. It is not an experimental study and therefore provides neither a validated treatment protocol nor comparative clinical outcomes.

Movement, forces and equilibrium

Kinematics describes movement-displacement, velocity and acceleration-whereas kinetics examines the forces and moments producing it.

A force is a vector defined by its magnitude, direction, sense and point of application. Under static conditions, forces and moments balance: $\Sigma F = 0$ and $\Sigma M = 0$. Under dynamic conditions, the resultant force produces acceleration according to $\Sigma F = m \cdot a$, while the resultant moment produces angular acceleration according to $\Sigma M = I \cdot \alpha$.

An oblique force can be resolved into a normal component F_n , producing compression or distraction, and a tangential component F_s , producing shear or translation.

Moment and lever arm

The ability of a force to produce rotation depends on its magnitude and perpendicular distance from the axis: $M = F \times d$.

For an unchanged force, increasing the lever arm therefore increases the moment. Conversely, halving the lever arm requires approximately twice the force to maintain the same moment.

In a dynamic orthosis, traction is mechanically maximal when perpendicular to the mobilised segment. If angle θ is measured relative to this perpendicular, the effective component becomes $F_c = F \cdot \cos \theta$ and the joint moment $M = d \cdot F \cdot \cos \theta$.

Effectiveness is approximately 100% at 0°, 87% at 30°, 71% at 45°, 50% at 60° and zero at 90°. As traction becomes more oblique, the corrective moment decreases and a shear component develops.

Stress and strain

Stress is force divided by the area over which it acts. Normal stress is defined as $\sigma = F_n/A$ and shear stress as $\tau = F_s/A$. With the same force, a smaller contact area therefore produces higher stress, supporting the need to distribute pressure under an orthosis.

Relative tissue strain is expressed as $\epsilon = \Delta L/L$. In a simplified linear model, the stress-strain relationship is $\sigma = E \cdot \epsilon$, where E is Young's modulus. The higher E , the stiffer the material.



Elastic deformation disappears when loading is removed. Plastic deformation persists after unloading. Failure occurs when the tissue's mechanical capacity is exceeded.

Time-dependent tissue behaviour

Biological tissues are viscoelastic: their response depends on load magnitude, duration and repetition.

Creep: under constant stress, deformation increases over time.

Stress relaxation: when deformation is held constant, stress progressively decreases.

Hysteresis: some energy is dissipated between loading and unloading.

Fatigue: repeated loading can cause cumulative damage even below the failure load.

Skin, tendons, ligaments and bone are not ideal materials: they are generally non-linear, anisotropic, heterogeneous and time-dependent.

Clinical message

The effect of a mechanical intervention never depends on force magnitude alone. It results from the combination of force + direction + point of application + lever arm + contact area + duration + repetition.

These principles support reasoning about traction direction, contact distribution, timing of loading and monitoring of tissue tolerance. However, they provide a general mechanical framework and cannot, by themselves, define a treatment protocol.

