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Biomechanics of Musculoskeletal Injury

IL Gitajn[1] and EK Rodriguez[2]
[1]*Harvard Combined Orthopaedic Surgery Residency Program, Massachusetts General Hospital*
[2]*Beth Israel Deaconess Medical Center, Department of Orthopaedic Surgery, Harvard Medical School*
United States

1. Introduction

Fracture as a result of traumatic injury is a major contributor to long-term disability and loss of work and is therefore an important health concern, as well as contributor to overall societal economic burden. Finklestein et al reported that the annual medical cost of traumatic injury in 2000 in the US was $80.2 billion and that the cost of productivity losses was $326 billion (Finklestein et al, 2006). A total of 1.5 million fractures occur each year, including 280,000 hip fractures and 500,000 vertebral fractures (Bouxsin et al, 2006). Because the human musculoskeletal system is a living organ with predominantly a mechanical role, physiology and engineering principles are critical for its study and understanding. Fracture and musculoskeletal injury occur when local stresses or strains exceed the ultimate strength of bones, tendons, ligaments and muscles. These tissues regenerate, heal, or fail to heal according to both mechanical and biological stimuli. This chapter will provide an overview of the biomechanics of musculoskeletal injury.

2. Acute injury and inflammation

Injury occurs when local stress or strain exceed the ultimate strength of bones and soft tissue. Since all tissues are to some degree viscoelastic, the rate at which energy is dissipated also contributes to the degree of tissue injury since tissue stiffness, which often defines failure modes, is dependent on rate of deformation. Unlike most materials, living tissues also respond to a traumatic event, not only with mechanical failure, but with an acute inflammatory response. This inflammatory response results in the sudden and extended release of inflammatory mediators, cytokines, and other factors that act, not only locally to define the injury and to initiate what ultimately will be the healing response, but also may have significant systemic effects, potentially resulting in severe pulmonary injury or end stage organ failure. Inflammatory cascades are initiated, not only in traumatized tissues, but also by pathogens, or other foreign irritants. In the setting of trauma, these inflammatory mediators are intimately associated with the healing process. They attract precursors for cell growth, and they modulate repair mechanisms. Inflammation also stimulates and increases the sensitivity of pain receptors, which serve a protective purpose, causing trauma patients to limit motion around the damaged tissue.

Inflammation is an acute immune response, designed to rid the organism of both the initial cause of cell injury and the consequences of such injury. In trauma, inflammation is triggered by pathogens, tissue necrosis, and foreign bodies. The inflammatory cascade is amplified by early recruitment of inflammatory cells, which in turn release further mediators. In the setting of trauma, the amount of inflammation is usually determined by the amount of energy transferred to the soft tissue and bone, the degree of contamination, and type of bacteria, if any, present, as well as patient factors, such as preexisting immunodeficiency, diabetes, or steroid use. The magnitude of the inflammatory response depends on the severity of the injury and the degree of vascularization of the tissue that is injured (Smith et al, 2008). Inflammation is likely initiated by cellular damage and subsequent leakage of intracellular contents, as well as by capillary damage, leading to blood flow into the site of injury and initiation of the injury hematoma.
Inflammation is primarily represented by four major events: vasodilatation, increased micro vascular permeability, cellular activation and adhesion of immune cells, and coagulation (Kumar et al, 2009). Vasodilatation and increased permeability of microvasculature permit extravasations of protein-rich fluid into tissues. This fluid consists of macrophages and monocytes, which release and stimulate cytokines and growth factors. Loss of fluid and increased vessel diameter lead to slower blood flow and vascular congestion (Kumar et al, 2009; Schroeder et al, 2009). Once leukocytes have been recruited to the site of injury, they are activated by intracellular components in the extracellular space, by proteins expressed on the surface of dead cells, or by cytokines.

2.1 Inflammatory mediators

There are several important mediators in inflammation, and a complete discussion is beyond the scope of this chapter. They can be categorized into cell-derived mediators, which may be sequestered into granules (histamine) or synthesized de novo (prostaglandins, cytokines), and plasma-derived mediators, which circulate as inactive precursors. Active mediators are produced in response to substances released from necrotic cells or microbes, and one mediator can stimulate the release of others. Platelets are an important source of cytokines and growth factors, and they are stimulated to release these cellular products during clotting, which occurs when platelets come in contact with collagen immediately after trauma is sustained (Diegelmann & Evans, 2004). There is increasing interest in the orthopedic community on the use of platelet enriched products as a therapeutic option for a variety of musculoskeletal conditions, ranging from tendon injury to bony nonunions (Hamilton et al, 2011; Mei-Dan et al, 2010; Sanchez et al, 2009). Histamine is present in mast cell granules and can be released in response to trauma, producing dilatation of arterioles and increased permeability of venules. Prostaglandins are a group cell derived mediators that can cause vasodilation, fever, and pain. The mechanism of NSAIDs' (non steroidal anti-inflammatory) anti-inflammatory action is by inhibiting cyclooxygenase, which is an enzyme that is critical in prostaglandin formation. Leukotrienes increase vascular permeability and cause chemotaxis and leukocyte adhesion.
Cytokines exert their effects by binding to specific cellular receptors and are thus able to regulate gene transcription and modify intracellular signally pathways, both locally and systemically. They have small molecular weight and are active in extremely low concentrations. They have overlapping functions, multiple targets, and pleiotrophic actions. TNFa and IL-1 are two important early pro-inflammatory cytokines. They affect a wide

variety of cells to induce fever, production of cytokines, endothelial gene regulation, chemotaxis, leukocyte adherence, and activation of fibroblasts. They are responsible for the systemic effects of inflammation, such as loss of appetite and tachycardia (Reikeras, 2010). IL-6 is another cytokine that appears to be critical in the inflammatory cascade in the setting of trauma. IL-6 levels are elevated 60 minutes after trauma (or surgery) and decline over days 2 to 5 after trauma. Importantly, the magnitude of IL-6 elevation after mechanical trauma can be used as a reliable marker for the magnitude of systemic inflammation and correlates with the risk of post-injury complications (Reikeras, 2010; Biffl et al, 1996; Pape et al, 2007). IL-6 appears to be responsible for regulating the acute phase response (Reikeras, 2010).

2.2 Systemic response

The inflammatory cytokines act locally, as well as systemically, and can lead to signs and symptoms similar to sepsis, including hypotension, fever, fatigue, anorexia, headache, activation of coagulation, and other systemic changes known together as the Systemic Inflammatory Response Syndrome (SIRS). This syndrome is most commonly seen in the setting of a serious bacterial infection and is initiated by circulating bacteria triggering an intense systemic inflammatory response. SIRS however does not require a setting of infection and can occur only as the result of injury and an inflammatory cascade. There is recent evidence that the systemic release of mitochondrial DNA and mitochondrial molecular patterns, which can occur with cellular breakdown in trauma, play a role in activating systemic inflammation in SIRS that is not a result of bacterial infection. Mitochondrial DNA and molecular patterns are similar to that of bacteria because they were likely derived from similar ancestors prior to the incorporation of mitochondria into human cells. Because of this similarity mitochondrial DNA and molecular patterns may trigger this intense inflammatory response by binding to the same immune receptors that recognize circulating bacteria (Zhang et al, 2010).

SIRS can be of severe consequence to the already debilitated trauma patient, resulting in pulmonary function collapse and organ failure. Careful consideration of timing is critical in the care of the trauma patients since further surgical intervention can worsen the inflammatory response. In severe polytrauma patients, it is often preferable to perform limited fracture stabilization, rather than definitive orthopaedic repair immediately, since surgery can function as a second traumatic event with a second wave of inflammatory cytokine release, which can augment the initial systemic inflammatory response to the trauma with increased potential to cause systemic disease including SIRS and ARDS (Second hit theory) (Reikeras, 2010; Pape et al, 2003; Sears et al, 2009). Hauser et al reports that SIRS is universal after traumatic injury and that the clinical presentation differs only in intensity (Hauser et al, 2010). One study showed that combined fracture and soft tissue injury caused higher levels of systemic inflammatory mediators (IL-6 and IL-10) than either fracture of soft tissue injury alone. The literature on SIRS and orthopedic trauma is extensive (Hardwood et al, 2005; Seibel et al, 1985; Scalea 2000; Olson, 2004; Schroeder et al, 2009; Sears et al, 2009; Weninger et al, 2007) with the femoral fracture being the primary model since it is a long bone fracture and is often most related with systemic and pulmonary collapse secondary to injury and surgery. Concern about the timing of definitive intramedullary fixation, which includes intramedullary reaming and further release of marrow contents and inflammatory mediators, is an ongoing debate in the orthopedic trauma community. It has been clear for several decades that early surgical

stabilization of long-bone fractures reduces pulmonary complications when compared to limb placed in a splint or skeletal traction. However, patients who are hemodynamically unstable, hypothermic, who have coagulation abnormalities or poor oxygenation due to traumatic lung injury have increased rates of acute lung injury after intermedullary reaming. If these conditions cannot be reversed with adequate resuscitation, these patients benefit from a protocol of damage control orthopaedics consisting of initial external fixation for transient stabilization followed by delayed definitive fracture fixation stabilization followed by delayed definitive fracture fixation (Bone & Giannoudis, 2011; Giannoudis et al, 2009; O'Toole et al, 2005, Hardwood et al, 2005; Sears et al, 2009; Pape et al, 2009; Pape et al, 2007; Pape et al, 2003).
Although inflammation is potentially harmful, with the ability to induce both local and systemic responses, it is also necessary to initiate the healing process. The inflammatory cells and proteins release growth factors and chemokines that recruit stem cells and other precursors and immune cells to the site of injury. These are then activated and stimulate others into becoming mitogenically active and proliferative. Even the hematoma and fibrin clot that occurs at the time of injury is important, likely providing a provisional structure for regenerative cells (Diegelmann & Evans, 2004). Studies demonstrate that when inflammation is limited, either in knockout mice or by pharmacological intervention, healing does not occur normally or is disrupted in time and sequence (Pape et al, 2007).

3. Bone material and structural properties

3.1 Introduction

Because the human musculoskeletal system is a living organ with predominantly a mechanical role, both physiology and engineering principles are critical for its study and understanding. The critical feature of any structural design is to consider what loads the structure must sustain and to adjust the overall geometry and the materials used to achieve the desired function. This is true in the musculoskeletal system as well.
The main function of the musculoskeletal system is to support and protect soft tissues and to assist with movement. Bones, muscle, tendons, ligaments and joints function to generate and to transfer forces so that our limbs can be manipulated in three-dimensional space. The musculoskeletal system also has a metabolic role in calcium handling, as well as hematopoiesis. To optimize function, bones must be rigid enough that they don't fail when loaded or demonstrate unnatural elastic behavior. They must also be elastic enough to absorb energy when loaded, but not so elastic that they are subject to plastic deformation (Seeman, 2003, 2006). The primary function of the musculoskeletal system is to manage applied load. The ability of a bone to resist fracture depends on the intrinsic properties of the material and the spatial distribution of bone mass (geometry and micro architecture) (Bouxsein & Karaski, 2006).

3.2 Material properties

Material properties characterize the behavior of materials comprising the tissue and to a first approximation, are independent of the size of the tissue. They are usually expressed in terms of the stress-strain relationship of the material. Stress is the amount of force applied per unit area, and strain represents the degree of deformation in response to a specific stress. Elastic deformation is the component of the stress-strain relationship in which the material

deforms as load is applied yet returns to its original shape when the load is removed. The slope of this curve is the elastic modulus or Young's modulus and it is a measure of stiffness. The stiffer the material is, the steeper the slope (the less it deforms under stress). Bone is an anisotropic material with a nonlinear stress-strain relationship that can be approximated as linear in its elastic region. When bone is loaded in the elastic range it absorbs the energy by shortening and widening in compression, lengthening and narrowing in tension, and then returning to its original length when unloaded (Chavassioux et al, 2007). Plastic deformation describes the condition in which some permanent deformation remains after the load is removed. With regards to bone, deformation in the plastic zone includes micro-cracks and disruption of collagen fibrils and its trabecular architecture. The anelastic modulus describes the slope of the stress-strain curve in the plastic range. Once the load exceeds the plastic deformation zone, the energy is dissipated in fracture or tissue failure. The yield point is the point at which elastic behavior changes to plastic, and it essentially describes the safe functional load. Subtle changes in density, which can occur with aging, disease, use and disuse, greatly change strength and elastic modulus (Browner et al, 2009; Bucholz et al, 2005).

Material properties of bone are generally separated into the material properties of the outer cortex and material properties of trabecular bone, which is found inside the cortex. These structures serve slightly different purposes and this is reflected by their material properties as well as the architecture. Bone is an anisotropic material; the stress-strain behavior differs with different directions of loading. Cortical bone is stronger and stiffer when loaded in the longitudinal direction than in the transverse direction. This is related to the orientation of bone microstructure (Browner et al, 2009). The orientation of orbicular architecture corresponds with the orientation of the principle stress sustained by the tissue (Huiskes, 2000). In less anisotropic bone, trabecular bone consists of cylindrical struts extending about 1mm before making connection with other struts, usually at right angles. In more highly anisotropic bone, trabeculi are more sheet-like than cylindrical, and they are longer and preferentially aligned in one direction (Currey, 2002). On a molecular scale, regions of bone loaded in tension tend to have their collagen fibers oriented longitudinally, while those loaded in compression tend to be oriented obliquely to transversely and collagen fibrils have been found to be oriented in the direction of the trabeculae (Rupple, et al, 2008; Chavassioux et al, 2007). Because of the anisotropic nature of bone, there is not a single value for elastic modulus and hardness of cortical or trabecular bone. This anisotropic nature will play an important role in bone resistance to failure or fracture.

Bone mineral content contributes to stiffness of bone at the expense of flexibility, and it also has an effect on bone toughness. As mineral content increases up to 65%, toughness increases, and as mineral content exceeds about 65%, toughness begins to decline (Seeman 2003; Xiaodu & Puram, 2003). Toughness is determined by the material composition and the ability of the microstructure to dissipate deformation energy without propagation of a crack. Energy can be dissipated by viscoelastic flow and by the formation of non-connected micro-cracks (Petterlik et al, 2006). Collagen cross-links are known to limit crack propagation, thus increasing bone toughness. Collagen structure is another important contributor to bone material properties. The triple helix of collagen and its cross-links confer strength in tension and are closely related to post-yield properties of bone, particularly bone toughness and ductility (Seeman & Delmas, 2006; Ruppel et al, 2008; Xiaodu & Puram, 2003). Water content also plays a role in relative stiffness and toughness of bone. The collagen network is very

sensitive to the condition of hydration (Seeman & Delmas, 2006). Dehydrated bone exhibits increased stiffness and decreased toughness (Xiaodu & Puram, 2003). The literature on documenting the mechanical properties of bone in various forms of loading is extensive (Turner et al, 1998; Choi, 1990; Morgan et al, 2003; Rho et al, 1997; Nyman et al, 2006; Bonfield, 1987)

Bone also exhibits viscoelastic behavior; bone strength depends on rate of loading and it exhibits creep and stress-relaxation. At higher strain rates, both ultimate strength and elastic modulus increase (Browner et al, 2009; Courtney et al, 1994; Bucholz et al, 2005; Currey, 2002). Under constant loads, bone will continue to deform or creep. If the strain is held constant, the stress decreases with time (relaxation). If cyclic loading is applied hysteresis (a phase lag in which the shape of the unloading curve is different from the shape of the loading curve), occurs leading to a dissipation of mechanical energy. More simply, some solid materials can flow slightly , but not indefinitely, and the rate of flow is proportional to the load being imposed but also inversely proportional to some function of time that the load has been imposed (Currey, 2002).

Bone also exhibits fatigue, in which loads below the yield point applied in succession progressively create a crack that grows until the material fails at a stress that is below the yield point. The fatigue resistance of a material depends more on limiting micro-crack growth than micro-crack initiation, and in bone, fatigue resistance also depends how quickly the material is able to restore micro-cracks, or heal. Micro-crack propagation is limited by bone heterogeneity and microstructural features, like cement lines around each osteon and the interface between loose and dense lamellae (Chapuriat & Delmas, 2009; Chavassioux et al, 2007). However, unlike inert materials, bone is able to sense accumulation of micro damage and to repair it. The phenomenon of fatigue is responsible for stress fractures, which are commonly seen in athletes, like runners, who do not provide frequently loaded bones with the opportunity to repair micro-damage (Hughes & Petit, 2010).

3.3 Structural properties

Structural properties of the musculoskeletal system, which characterize the tissue in its intact form, also play a critical role in managing applied loads and particularly in transferring stress through the skeletal system. This takes into account the material properties of each type of tissue in the structure, as well as the geometry and architecture of the system. Overall strength of the system depends on the size and shape of the bone (cortical thickness, cross sectional area and moment of inertia), the micro-architecture of the bone (cortical porosity, trabecular morphology), and the amount of accumulated damage.

Moment of inertia is a measure of how the material is distributed in the cross-section of the object relative to the load applied to it, and moment of inertia can be used to predict the resistance of the structure to bending and deflection (Bucholz et al, 2005; Bouxsein & Karasik, 2006).

$$\text{Moment of inertia} = P\left(R^4 - r^4\right)/4$$
$$R = \text{cortical outer diameter};\ r = \text{cortical inner diameter} \quad (1)$$

Since moment of inertia is proportional to the diameter of the structure to the 4th power (Browner et al, 2009) small increases in external diameter of a long bone can markedly improve its resistance to bending and torsional loading (Bouxsein & Karasik, 2006). Resistance to compressive loading depends on the cross sectional area of bone; resistance to

bending and torsional loads involves distributing bone material far from the neutral axis of bending or torsion (generally this axis is near the center of bone) (Bouxsein & Karasik, 2006). This is highly relevant for understating changes in bone properties with aging. Osteoporosis as a result of aging, not only results in decreased mineral bone content, but aging causes a architectural remodeling which affects the moment of inertia of bone. Geriatric patients have long bones characterized by an increase in external diameter and a larger increase in internal diameter, resulting in a thinner cortex (figure 1). The increased inner diameter (and thinner cortex) results in significant decreases in bone bending resistance since moment of inertia is directly related to ($R^4 - r^4$). This is countered, to some degree by the increase in outer cortical diameter.

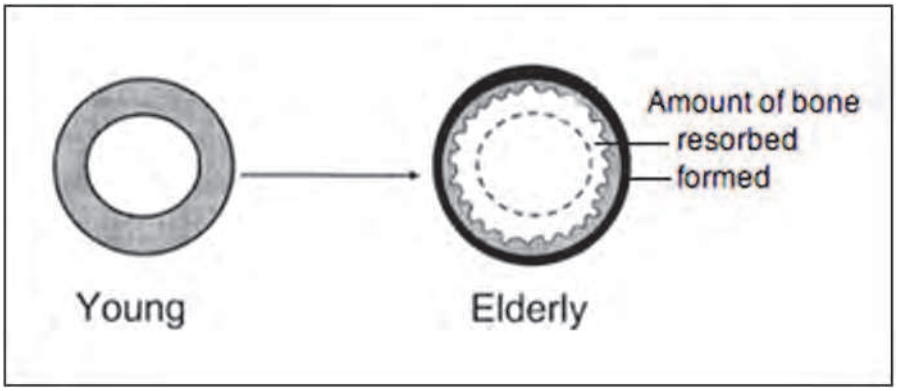

Fig. 1. Change in bone cortical diameter with age. Figure adapted from Seeman, 2003, with permission of Elsevier Limited.

Geometry is difficult to discuss in general over the entire skeleton because it is not uniform; skeletal structure and geometry is specific to the needs of each anatomical region. For example, long bones are needed for loading and movement, and rigidity in these bones is therefore favored over flexibility (Seeman & Delmas, 2006; Chavassioux et al, 2007). By shifting the cortical shell outward from the neutral axis, the long bones have increased bending strength. External and internal contours differ at each point along and around the shaft, reflecting local modeling and remodeling in response to regional loading needs (Chavassioux et al, 2007). The reverse is true in the vertebrae, where ability to deform in response to loading is favored over stiffness. Vertebral bodies with large volume of trabecular bone function more like springs than levers. Interconnecting trabecular plates achieve lightness and favor structural flexibility over stiffness (Seeman & Delmas, 2006; Chavassioux et al, 2007). Additionally the diameter and thickness of bones is different, depending on the types of stresses that are sustained by that bone. For example, the femoral neck adjacent to the shaft is elliptical, with the longer diameter in the superior-inferior direction with greater cortical thickness inferiorly. These geometrical features minimize bending. Near the femoral head, stresses are mainly compressive and the geometry reflects this. The femoral neck is more circular and largely trabecular, with a cortex of similar thickness around its perimeter (Seeman & Delmas, 2006). Later sections will elaborate on biomechanical changes that occur with age and how this affects propensity for musculoskeletal injury.

3.4 Remodeling

In 1982 Julius Wolff published a paper on bone remodeling, defining a phenomenon that would become known as Wolff's law--that bone changes external geometry and internal architecture in response to stresses acting on it (Wolff, 1986). Wolff's law has been quoted in numerous ways through the years and referenced for support whenever the argument of stress modulated bone remodeling was being made. However, Wolf's law is not a law in the quantitative sense but rather an insightful observation. There is no known growth law for bone or any other musculoskeletal tissue that is universally applicable or demonstrated. It is also unclear if bone remodeling is a stress- or strain-governed phenomenon.

During the remodeling process, osteoclasts (bone resorption cells) remove old bone tissue by resorption, and osteoblasts (bone forming cells) create new bone tissue. It is understood that bone is remodeled to meet its mechanical demands. There is evidence that micro damage initiates bone remodeling and that fracture repair is a form of load-induced bone remodeling in which damage serves as trigger (Chapuriat & Delmas, 2009; Burr et al, 1985; Mori & Burr, 1993). Stress fractures are often localized radiographically when patients complain of limb pain, and a radiograph demonstrates a reactive response or fracture callus that illustrates the remodeling process initiated by the injury. The principles of remodeling and bone fracture healing with callus often reflect the need to redistribute stress at the site of healing. A large callus that increases the cross sectional areal of the bone at the site of a transverse shaft fracture serves a means of increasing the moment of inertia and decreasing the bending stress sustained at the fracture site.

Evidence for exercise-induced osseous remodeling in adults is less clear. Data from intervention randomized control trials is limited. Follow-up times have been short, the quality of the conduction of intervention and reporting of outcomes has been poor, and there has been a lack of reporting on the specific exercise characteristics that are effective (Korpelainen et al, 2006; Bonaiuti, 2004). However, adaptation to loading in children and adolescents is well documented, and these changes in bone density and geometry persist into adulthood. Exercise, particularly weight-bearing impact exercises, in prepubertal boys increases estimates of bone strength at loaded sites, likely due to thicker cortices (Nikander et al, 2010; Nara-Ashizawa et al, 2002). Young tennis players have increased cortical thickness and increased cortical drift in the perosteal direction in their playing arm compared with their non-dominant arm. However, in middle-aged subjects, tennis did not stimulate cortical drift in the periosteal direction. In middle-aged subjects cross-sectional areas of the radius were actually smaller, suggesting that unilateral use of the arm after the third decade of life suppresses age-related changes in bone geometry since normally there is increased endocortical area and slower expansion of periosteal area resulting in decreased cortical thickness (Nara-Ashizawa et al, 2002). There is some evidence that exercise can increase bone mineral density (BMD) in post-menopausal women, particularly after one year or longer. The type of exercise and the amount of improvement is somewhat contested. A few studies, however, suggest that resistance training and low- to moderate-impact exercises are most effective. However the gains in BMD are generally small (1-2%) (Nikander et al, 2010; Korpelainen et al, 2006; Bonaiuti, 2004). Exercise has been shown to result in up to a 50% reduction in fracture incidence, but a large component of this reduction is likely due to improved muscle function and balance, combined with the small 1-2% increase in BMD (Nikander et al, 2010).

The cellular mechanism for remodeling control is a focus of research interest, but the details are still largely unknown. Osteocytes appear to be the primary mechanosensors that begin the remodeling cascade. There is evidence that pressure gradients within the bone matrix

lead to interstitial fluid flow in the lacunar-canalicular system, which activates mechanosensory osteocytes that reside in lacunae. The osteocytes then transmit load-provoked signals via canaliculi and gap junctions (Chen et al, 2010; Ulstrup, 2008). There is evidence that osteocyte death is associated with remodeling as well (Seeman & Delmas, 2006). Death of cells likely creates biochemical and chemotactic signals, which indicate presence of damage and its location. Regions of micro damage contain apoptotic osteocytes whereas quiescent zones do not (Seeman & Delmas, 2006; Hughes & Petit, 2010).

3.5 Mechanics of bone regeneration

Under most circumstances bone is able to regenerate its baseline mechanical properties after sustaining a fracture. However, the mechanical environment is critical in establishing tissue formation patterns during fracture repair. There are two forms of bone healing: direct or primary healing and indirect or secondary healing, which occur depending on the mechanical environment. Direct healing occurs when the fracture is subjected to surgical fixation with absolute stability, fixation with absolute stability, with no interfragmentary motion or strain with no interfragmentary motion or strain. This direct healing is achieved by interfragmentary compression, most often achieved technically during surgery with lag screws and/or compression plates. In this setting, bone heals via intramembranous ossification without development of a fracture callus. This is most often applied to peri-articular fractures where perfect anatomic reduction is necessary for an optimal functional outcome. Indirect healing occurs when the fracture is subjected to relative stability, or when there is some degree of interfragmentary motion or strain. Bone heals with development of a fracture callus, which changes the mechanical properties and the geometry of the fracture site. This often produces optimal biological conditions for healing.

Interfragmentary strain theory, pioneered by Perren in 1979 is the basis for our understanding of how the mechanical environment impacts tissue differentiation in a fracture gap (Figure 2). He theorized that the magnitude of interfragmentary strain determines subsequent tissue differentiation of fracture gap tissue. Each tissue has different strain tolerances, and applied interfragmentary strain must be smaller than the strain tolerance of a tissue for it to form. According to Perren, strains below 2% permit direct bone formation (direct fracture healing), strains below 10% allow cartilage differentiation and subsequent endocondral ossification (indirect fracture healing), and strains between 10% and 100% lead to granulation tissue formation and non-union. Perren believed that differentiation of initial fracture gap tissue would stiffen the fracture gap leading to lower interfragmentary strain, allowing differentiation to the next stiffest tissue. (Perren, 1979; Perren, 2002; Isaksson et al, 2006).

Carter and Blenman supplemented Perren's theory with the idea that, in addition to strain magnitude, both the type of mechanical stimulus (cyclic, compressive, tensile or shear) and the degree of vascular supply would affect tissue differentiation. Prendergast et al later developed a different mechanoregulation concept that proposed two biophysical stimuli, shear strain in the solid phase and fluid velocity in the interstitial fluid phase. According to this concept, bone formed only when both stimuli were low enough. However, none of these models are flawless, and clinical results suggest that these theories are correct in the extremes, where they are similar: low strain leads to bone formation, and high strain leads to fibrous non-union. (Carter et al, 1998; Carter et al, 1988; Prendergast et al, 1997; Isaksson et al, 2006).

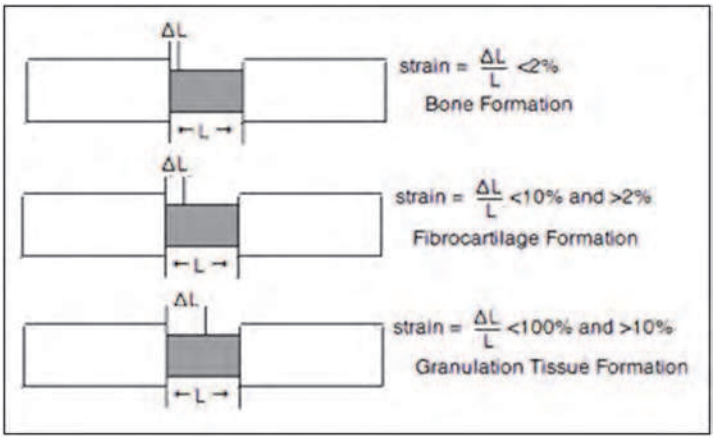

Fig. 2. Interfragmentary strain theory. The formation of tissue type based on strain at the fracture gap

4. Geriatric biomechanics

4.1 Osteoporosis

Osteoporosis is defined as a "systemic skeletal disease characterized by low bone mass and micro-architectural deterioration of bone tissue, leading to enhanced bone fragility and a consequent increase in fracture risk "(Alexeeva et al, 1994). In the US over 1.5 million fractures occur each year, including 280,000 hip fractures, and these numbers are expected to double or triple in the coming decades due to the aging population (Bouxsein & Karasik, 2006). There are several components of whole bone strength that change over time, including the intrinsic properties of the materials that form bone, the amount of bone (ie mass), and the spatial distribution of bone mass (ie geometry and microarchitecture) (Bouxsein & Karasik, 2006). The biggest challenge is determining the effects of these changes and identifying which change is most important in the development of osteoprosis.

4.2 Bone mechanical properties

Whole bone strength declines dramatically with age. Changes that occur in cortical, as well as trabecular bone collectively lead to decreased bone strength and increased risk of fracture. Between 30 and 80 years of age, elastic modulus of cortical bone decreases by 8%, bone strength decreases by 11%, and toughness declines by 34% (Bouxsein & Jepsen, 2003). All of these changes result in mechanical failure or fracture as a result of lower energy traumatic events. The specific changes that contribute to these events are a topic of investigation. It is clear that there is a reduction in overall bone mass with age. It is thought that thinning of cortical bone and increased porosity are major contributors to loss in stiffness, strength, toughness, and resistance to propagation of cracks (Silva, 2007; Seeman 2006). Studies have shown that there is a four-fold increase in cortical bone porosity from 20 to 80 years of age (Brockstedt et al, 1993). The elastic modulus of cortical bone in a longitudinal direction decreases significantly with increased porosity (Schaffler & Burr, 1988; Currey, 1988; Dong & Guo, 2004). Other factors that may contribute to decreased toughness include loss of bone mass, increased mineralization or development of

hypermineralized regions, accumulation of micro damage, decreased integrity of collagen, and changes in collagen crosslinks (Xiaodu & Puram, 2003).
The changes in mechanical properties of trabecular bone are even more pronounced. Between 30 and 80 years of age, elastic modulus of trabecular bone decreases by 64%, strength decreases by 68%, and toughness decreases by 70% (Bouxsein & Jepsen, 2003). These changes are likely due to loss of trabecaular plates and connectivity, as well as micro damage. Studies have shown that loss of connectivity in trabecular plates produces a greater deficit in bone strength than thinned plates that continue to be well connected (Seeman & Delmas, 2006; Silva, 2007; Chavassioux et al, 2007).

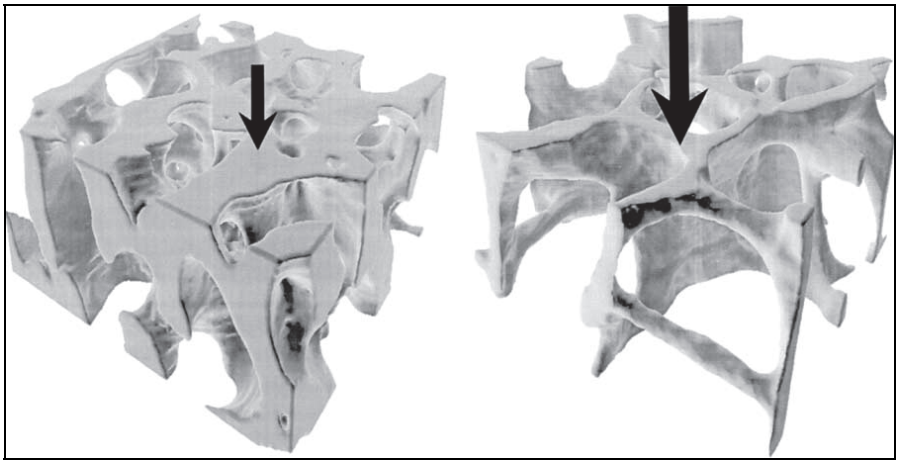

Fig. 3. Loss of trabecular bone mass and decreased trabecular connectivity ocurs with increasing age. Figure from Seeman, 2003, with permission from Elsevier Limited.

4.3 Bone geometry

The overall size and shape of bones play important roles in their mechanical behavior. Microarchitectural changes in trabecular bone, such as decreased number of trabecular plates and decreased connectivity between plates, appear to play a large role in decreased strength of trabecular bone. Decreases in bone mass and changes in distribution of bone mass also appear to play a large role in overall bone strength (McCreadie & Goldstein, 2000; Kreider & Goldstein, 2009). It is well established that endosteal expansion (increase in inner cortical radius due to loss of cortical bone) and perioesteal expansion (increase in external cortical radius due to deposition of new bone on the external surface of bone cortex) both occur, but that endosteal expansion exceeds periosteal expansion (Figure 1). This excess leads to age-related deceases in cortical thickness but increases in bone outer diameter. Decreased cortical thickness contributes to the decreases in strength, elastic modulus, and toughness of bone. However, the greater diameter increases moment of inertia and increases structural resistance to bending and torsional loads, which may offset decreases in cortical thickness and bone mineral density that occur with age. This effect explains how bone mineral density can decrease while bending resistance may not (McCreadie & Goldstein, 2000; Beck et al, 2000; Bouxsein et al, 2006; Silva, 2007; Seeman & Delmas, 2006; Mayhew et al, 2005).

Bone is known to be highly anisotropic at baseline and is strongest in the direction of habitual loading. There is emerging evidence to suggest that in hip-fracture patients, bone is more anisotropic and more highly oriented in the direction of habitual loading than control subjects, occurring at the expense of strength in other directions. One study examined specimens from hip fracture patients and un-fractured controls, and controlled for bone volume. The study found that hip fracture specimens of the same bone volume were more highly organized in the direction of habitual loading. This increased anisotropy leads to a reduced ability to withstand off-axis impact, during a fall in a direction different from the direction of habitual loading, such as a sideways fall. In these patients, bone reorganization may be overcompensating for the low mass status by increasing the degree of anisotropy so that strength of the bone is only maximized in the frequently loaded direction (Kreider & Goldstein, 2009; Ciarelli et al, 2000; McCreadie & Goldstein, 2000).

The rate of bone remodeling may also play a role in development of osteoporosis. During growth the balance between bone that is removed and bone that is formed is positive (more bone is added than removed). Once skeletal maturity is reached, this reverses and the balance becomes negative (more bone is removed than is added in the remodeling process). In general, the rate of remodeling, and therefore the rate of bone loss, is extremely slow later in life. However, there is evidence to suggest that estrogen deficiency increases the rate of remodeling and there may be other factors that modulate remodeling rate. It is possible that bone loss is driven more by increased rate of remodeling than by magnitude of bone loss during each remodeling event (Seeman & Delmas, 2006; Xiaodu & Puram, 2003). Another possible mechanism through which bone remodeling contributes to osteoporosis is through increasing dysfunction of mechanoreceptors, which drive the remodeling process. This could contribute to bone loss and could interfere with remodeling in response to micro damage or in response to changes in loading (Kreider & Goldstein, 2009).

4.4 Bone mineral density

Bone mineral density (BMD) is the attribute currently used in clinical practice to diagnose osteoporosis and to monitor efficacy of interventions. Dual-emission X-ray absorptiometry (DXA) is used to measure BMD clinically. BMD is bone mineral content (BMC) (measured as the attenuation of the X-ray by the bones being scanned) divided by area of the site being scanned. Osteoporosis is diagnosed by determining how many standard deviations the BMD of the patient is below the mean BMD of a healthy thirty year-old. Any BMD that is greater than two and a half standard deviations below the mean thirty year-old BMD is considered osteoporotic.

BMD explains a significant portion of the risk of osteoporotic fracture and correlates with bone strength. BMD is a strong predictor of fracture risk; risk of fracture increases 50-150% with each standard deviation decrease in bone mass as measured by DXA (McCreadie & Goldstein, 2000; Kreider & Goldstein, 2009; Bouxsein et al, 1999). However, it is clear that there are other factors that contribute to fracture risk. Studies have demonstrated that there is a significant overlap in BMD between osteoporotic individuals and healthy individuals who have not experienced osteoporotic fracture. The risk of fracture of the hip or forearm in a 75 year-old is 4-7 times that of a 45 year-old with an identical BMD. Risk for hip fracture actually doubles for each decade of age increase even after adjusting for bone density (Beck et al, 2000; Ruppel et al, 2008; Degoede et al, 2003). Additionally, current therapies are able to, at best, increase bone density by 10%, but the risk of fracture decreases by a much larger

extent (McCreadie & Goldstein, 2000). The specific non-BMD factors that explain this discrepancy are not known.

BMD is used clinically because it represents a non-invasive, relatively inexpensive way, to predict fracture risk. It indirectly reflects bone geometry, mass, size, and mineralization. However, DXA does not provide information on cortical vs. cancellous density, 3D geometry, trabecular architecture, microstructure or strength parameters. It functions as a surrogate for these attributes, which are difficult to measure non-invasively (Kreider & Goldstein, 2009; Bouxsein et al, 1999).

4.5 Bisphosphonates

Bisphosphonates are a class of drugs that are commonly used to manage osteoporosis. They function by inhibiting bone resorption by osteoclasts, which occurs during remodeling. They mimic the structure of pyrophosphate and are incorporated into bone. They are then ingested by osteoclasts and ultimately result in osteoclast cell death. During bisphosphonate treatment bone remodeling rate is slower and there are a fewer number of osteoclast-induced excavation sites each with decreased depth, leading to slower bone loss. Fractures are less frequent but not eliminated in patients taking bisphosphonates (Seeman & Delmas, 2006). Maximum fracture risk reduction occurs in the first year of treatment. Observed fracture risk appears to be at least twice as large as would be expected from changes in BMD alone (Ruppel et al, 2008).

In recent years there has been some controversy with regard to safety of prolonged bisphosphonate administration. Several case series initially described cases of "atypical" subtrochanteric and diaphyseal femur fractures and suggested that the risk may be increased in long-term users of bisphosphonates (Black et al, 2010; Glusti et al, 2010; Capeci & Tejwani, 2009). Unique clinical features of these fractures in the literature include prodromal pain for weeks to months prior to fracture, complete absence of precipitating trauma, and bilateral fracture (either simultaneous or sequential) in some. Distinctive radiographic features include presence of a stress reaction on the affected and/or unaffected side, transverse or short oblique pattern (in contrast to the more common spiral fracture), thick femoral cortices, and unicortical breaking (Black et al, 2010; Glusti et al, 2010; Nieves & Crosman, 2010; Singer, 2011). The theory behind this concern is that long-term bisphosphonate use with prolonged suppression of bone turnover may lead to accumulation of micro damage due to impaired remodeling. It has also been suggested that long-term bisphosphonate use could create a more homogenous tissue with BMD more similar throughout, and this may offer less resistance to propagation of cracking (Glusti et al, 2010; Seeman & Delmas, 2006; Rupel et al, 2008). Fracture patterns and cortical thickening are reminiscent of osteopetrosis and fractures that occur in ostepetrosis in the subtrochanteric area (Armstrong et al, 1999; Golden & Rodriguez, 2010; Tolar et al, 2004; Singer, 2011). Osteopetrosis is a congenital malfunction of the osteoclast resulting in severe brittle and dense bone. There are several retrospective cohort studies that indicate that there is a correlation between atypical subtrochanteric and diaphyseal femur fractures and use of several bisphosphonates (Vestergaard et al, 2011; Lenart et al, 2009). However, it is difficult to determine whether this correlation is confounded by the fact that those taking bisphosphonates, particularly long-term, have significant osteoporosis that may account for these atypical fractures. Data from three large placebo controlled, randomized control trials have indicated that there is no association between bisphosphonate use and atypical

subtrochanteric or diaphyseal femur fracture. Based on these three studies, it is likely that these subtrochanteric and femoral shaft fractures may be related to the underlying osteoporosis, which was the reason for long-term bisphosphonate use (Black et al, 2010; Nieves & Crosman, 2010; Rizzoli et al, 2010) or to an additional metabolic predisposition yet to be dignosed. However, confidence intervals in these studies were high due to the small number of events, and, although, one study followed patients for ten years, it is possible that this is not long enough to observe an effect. Additionally, it has been suggested that these fractures are associated with bisphosphonate use in a subset of patients, like those taking steroids or proton pump inhibitors. Future studies will investigate these possibilities and bisphosphonates remain a valuable tool in the standard of care of the osteoporotic patient.

5. Fracture mechanisms

Injury patterns sustained in trauma can often be inferred from bone radiographs after trauma with certain confidence and consistency (Linnau et al, 2007; Clare, 2008; Mubarak et al, 2009; Arimoto & Forrester, 1980; Browner et al, 2009). Knowledge of patterns of injury attributed to specific modes of trauma can be used to predict associated injuries, since not all injuries are obvious at presentation. This knowledge also serves to develop or to improve safety features and equipment.

The magnitude, type and direction of forces, as well as material properties of bone and surrounding structures, dictate the fracture pattern to a certain degree. Severity of injury is determined by peak forces and moments resulting from the impact and the tissues' resistance to injury (DeGoede et al, 2003). The greater the energy absorbed by the bone, the more severe the fracture and the more likely that comminution and displacement will occur. Tissues surrounding bone, including muscle, tendons, ligaments, fat and skin, can affect fracture pattern by absorbing some of the load energy and also by creating additional load. The main factors that affect the load at which bone fails include bone geometry, bone material properties, load application point, load direction and the rate of load application (DeGoede et al, 2003). The main load bearing structure in bone is the cortex, which is denser, has greater volume and mass, and is in a location that makes it more capable of sustaining large loads. Trabecular bone largely functions to direct stresses to cortical bone. Multiple injuries can be caused by the same mechanism because forces can be transmitted along the entire length of a bone or through several bones, causing damage anywhere along the way.

5.1 Simple fracture patterns

There are a limited number of loading modes that bone can be subjected to, and these result in predictable fracture patterns. Complex fracture patterns occur when multiple loading modes and directions are applied during the same event. Loading modes include tensile loading, compressive loading, shear loading, bending load, and loading in torsion. Bone is weakest in tension and strongest in compression. When bone is loaded in tension it tends to fracture along a transverse plane that is approximately perpendicular to the direction of loading. When undergoing a compressive load, bone will fail secondary to shear stress since shear strength of bone is much less than compressive strength. During compressive loading, shear stresses develop at a plane that is approximately 45 degrees from the long axis of the bone, and it is along this oblique plan that bone fails. Max shear stress is approximately one half of the applied compressive stress. Bending is essentially a combination of tensile and compressive loading. When bone is undergoing bending, high

tensile stresses develop on the convex side and high compressive stresses develop on the concave side. A transverse fracture is initiated on the tensile side, and two oblique fractures occur on the compressive side, creating what is referred to as a butterfly fragment. Fracture secondary to torsion usually begins at a small defect at the bone surface, and then the fracture follows a spiral pattern along planes of high tensile stress, since bone is weakest in tension (Browner et al, 2009; Bucholz et al, 2005; Canale, 2002). It is a worthwhile exercise for a traumatologist to carefully look at a radiograph after trauma and to recreate the mechanism of fracture based on the fracture pattern. More complex and comminuted fracture patterns are essentially a combination of these simple patterns (Browner et al, 2009; Bucholz et al, 2005; Canale, 2002). Materials properties of bone can be approximated as isotropic when load is delivered at a high rate.

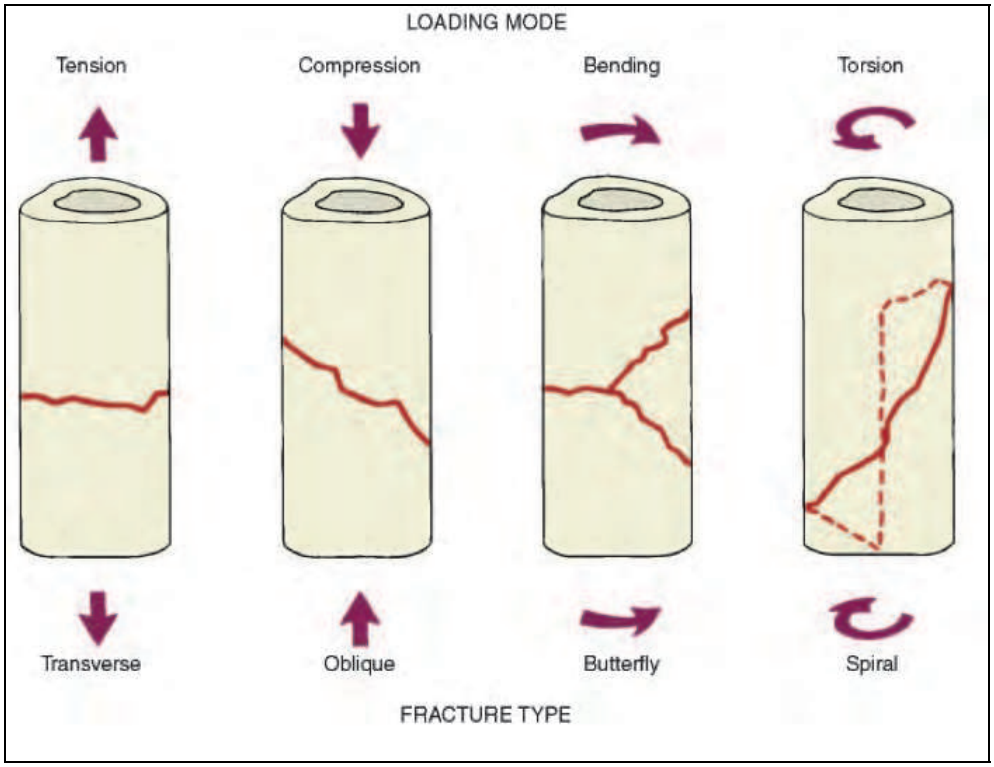

Fig. 4. Simple fracture patterns which occur as a result of loading mode. Figure from Browner et al, 2008, with permission from Elsevier Limited.

5.2 Fall

Fall is an important source of musculoskeletal injury and accounts for 87% of fracture in older adults (DeGoede et al, 2003). The two most common injuries secondary to fall are hip fractures and upper extremity fractures, and in some instances, they are related in that impacts at the wrist have been shown to modulate or lessen impacts at the pelvis during lateral and forward falls. This requires rapid reaction and movement times, as well as arm

muscle strength, all of which decrease with age to some degree (DeGoede et al, 2003). One study measured reaction time of young and elderly women and found that the typical elderly female is able move her hands quickly enough to break a forward fall, but not a sideways fall, while young women are able to break both types of fall (Robinovitch et al, 2005).

5.3 Fall on outstretched hand

Fall on outstretched hand is a classic mechanism of injury leading to fracture of the scaphoid bone of the hand, fracture of the distal ulna and radius, fracture-dislocation of the elbow, fracture dislocation of the shoulder and fracture of the clavicle. This injury mechanism accounts for approximately 90% of fractures at the distal radius, humeral neck, and supracondylar elbow region (Robinovitch et al, 1998). During a fall on a stiff surface, hand contact force occurs in two stages: the first is a high-frequency peak load which corresponds to a large deceleration of arm mass, which occurs at the wrist at the moment of impact; the second is a low-frequency oscillation with a lower peak force, which is due to deformation of the shoulder spring (Figure 5) (Chiu & Robinovitch, 1998). Increases in body mass more strongly increase the peak magnitude of the low-frequency component, and increases in fall height more dramatically increase the high-frequency component (Chiu & Robinovitch, 1998).

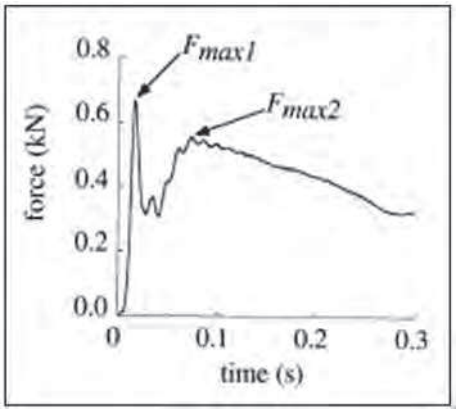

Fig. 5. Impact response of the body during a forward fall onto the outstretched hand. Measures of hand contact force during this event show a high-frequency transient (with associated peak force Fmax1) followed by a lower-frequency oscillation (with associated peak force Fmax2). Figure from Chiu et al, 1998 with permission from Elsevier Limited.

The fracture pattern depends on the force magnitude, on how force is distributed across the bones of the hand, and on how it is transmitted to other upper extremity structures. The magnitude and distribution of contact force during a fall also depends on the configuration of the body at impact and on the soft tissue thickness over the palm region (Choi & Robinovitch, 2010). The weakest area on the palm is over the scaphoid and lunate, which articulate directly with and transmit force to the distal radius. A fall with peak force localized to this area is most likely to result in fracture.

Understanding the mechanics of an injury helps to develop preventative measures. In one study patients were able to learn to reduce the impact force applied to the distal forearm by 27% by slightly flexing their elbows and reducing the velocity of the hands relative to the torso (DeGoede & Ashton-Miller, 2002). Another study showed that a 5mm foam pad reduced peak pressure and peak force by 83% and 13% respectively (Choi & Robinovitch, 2010), which can represent the difference between a fracture and isolated soft tissue damage. Additionally weight plays a role in degree of loading during a fall. Peak pressure was 77% higher in individuals with high body mass index (BMI) when compared to low BMI participants (Choi & Robinovitch, 2010). In contrast to what we see in the hip, having high BMI is not associated with increased thickness of soft tissue in the hand, and therefore the extra body mass contributed to the total force of the fall without providing extra tissue to absorb the energy.

5.4 Femoral neck fracture

Hip fracture or femoral neck fracture is a significant source of morbidity in the elderly population, and 90% of such fractures are due to fall from standing (Robinovitch et al, 1997; Parkkari et al, 1999). Hip fracture in the elderly is associated with a 20% chance of death and a 25% risk of long-term institutionalization (Parkkari et al, 1999). Changes that occur with aging in the material properties of bone play a significant role in femoral neck fracture; however, the mechanics of the fall (direction, location of impact) are critical as well. Although 90% of hip fractures are due to a fall, only 1% of falls actually result in hip fracture, which is surprising from a biomechanical perspective because the energy available during a fall from standing often exceeds that required to fracture both elderly and young proximal femurs (Robinovitch et al, 2000). Mitigating factors can be many.

The femoral neck undergoes constant bending loads during normal weight-bearing activities. Compressive force through the femoral head can range from 4-8 times the body weight during normal activities and this force acts through a significant moment arm (the length of the femoral neck), which causes large bending loads on the femoral neck (Browner et al, 2009). In normal gait the greatest stresses occur in the subcapital and mid-femoral neck regions. Within these regions maximum compressive stresses occur inferiorly where the cortex is thick and smaller tensile stresses occur superiorly where the cortex is thinner (de Baker et al, 2009). Sideways falls with impact to the greater trochanter are the events most directly related to hip fracture in older adults (Liang & Robinovitch, 2010; Parkkari et al, 1999; Courtney et al, 1994). The femoral neck is weakest when the posterolateral aspect of the greater trochanter is impacted. During a sideways fall on the greater trochanter, the stress state is reversed from normal ambulation and the greatest compressive stresses occur in the superior femoral neck while the smaller tensile stresses occur in the inferior region (Figure 6) (de Baker et al, 2004). Mayhew et al showed that the superior cortex of the femoral neck is significantly thinner in older than younger individuals, while the inferior cortex is significantly thicker in older than younger individuals (Mayhew et al, 2005). Therefore, during a sideways fall, which is more frequent in the elderly, the large compressive stress occurs in the superior cortex, which is thinner and more likely to fail in the elderly. Multiple studies have suggested that proximal femur fractures are typically initiated by a failure in the superior aspect of the femoral neck, followed by a failure in the inferior aspect of the femoral neck (de Baker et al, 2009; Lotz et al 1995; Mayhew et al, 2005). Wang et al showed that subjects with a longer moment arm in the context of a sideways fall increases the force

applied to the hip and predisposes the subject to a hip fracture. Hip axis length and neck-shaft angle both contribute to the moment arm of the hip and both have been independently shown to predict hip fracture (Wang et al, 2008; Leslie et al, 2009; Patron et al, 2006; Crabtree et a, 2002). A fall onto the greater trochanter may also generate an axial force along the femoral neck, resulting in an impaction fracture. Additionally, investigators have reported that the lower extremity externally rotates during a fall and that, at the extremes of external rotation, the femoral neck impinges against the posterior acetabular rim. The acetabular rim then acts like a fulcrum to concentrate the stress experienced by that region at time of impact (Koval & Zuckerman, 1994).

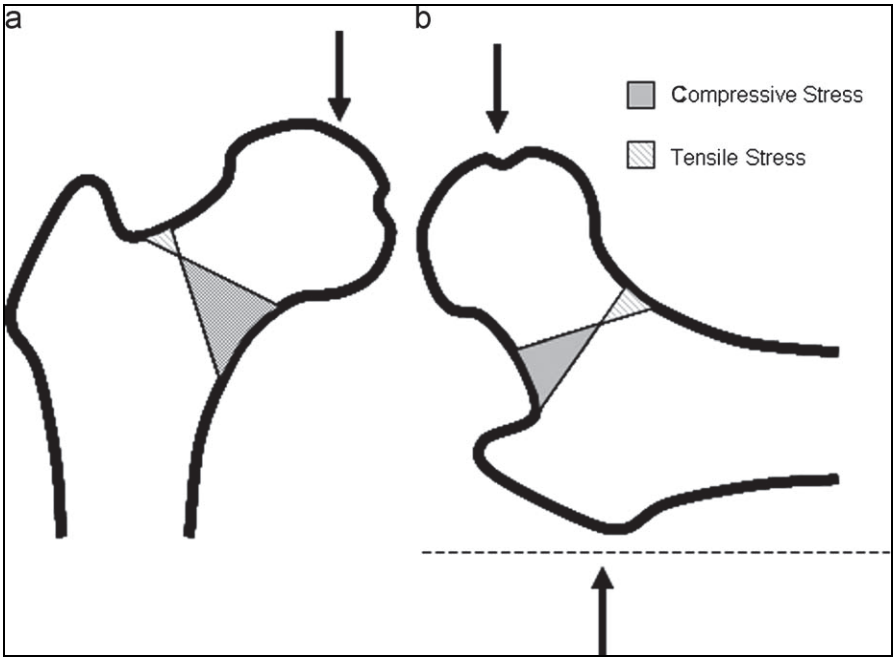

Fig. 6. The magnitude and nature of the stresses on the femoral neck differ depending on the applied load. For example in (a) walking: the inferior surface tends to be subjected to a large component of compressive stress, while the superior surface is subjected to a smaller tensile stress and (b) sideways fall on the greater trochanter: the inferior surface tends to be subjected to a small tensile stress, while the superior surface is subjected to a larger compressive stress. Figure from de Bakker et al, 2009 with permission from Elsevier Limited.

Since only a small fraction of falls actually result in fracture and the energy available in a fall is sufficient to fracture the proximal femur, there are mitigating factors that affect the actual impact forces. Some of these include soft tissue properties and body positioning at the time of impact. Energy of a fall can be dissipated by contracting muscles; this contraction is likely done more effectively in younger patients than older patients with slower, weaker muscles (Koval & Zuckerman, 1994). Substantial energy can also be absorbed by skin and fat overlying the hip region (Robinovitch et al, 2000). Peak femoral impact force actually

decreases in a linear manner with increasing soft-tissue thickness at a rate of approximately 79 N per 1mm change in thickness (Robinovitch et al, 1995), and peak pressure over the greater trochanter averaged 266% higher in low BMI participants than in high BMI participants in another study (Choi & Robinovitch, 2010). Additionally, there are actions that fallers can take to moderate the force applied directly to the femur. Falling techniques can be taught to geriatric patients by physical therapists. In one study young subjects were able to impact the outstretched hand and pelvis near-simultaneously during an unexpected fall which distributed the body's impact energy (Robinovitch et al, 2000). Fallers can also produce "energy absorbing" work during descent, which occurs by eccentrically contracting lower extremity muscles, which increases the vertical component of foot reaction forces resulting in decreased downward acceleration (Robinovitch et al, 2000). Mats as thin as 1.5cm have been shown to decrease peak hip impact force by 8% and thicker mats have a greater effect (Liang et al, 2006). Ultimately, these modifiable factors, which diminish the peak impact force, are critical because they represent ways that hip fracture can be reduced or prevented.

5.5 Motor vehicle collision

Motor vehicle collision is a common source of polytrauma, injuring more than 5 million people every year (Peterson et al, 1998). Generally, injuries are sustained when the vehicle rapidly decelerates while the vehicle occupant continues to move at previous speeds. When the body absorbs energy beyond its tolerance fracture or injury occurs. Since bone and soft tissue resistance to injury decreases with age, elderly vehicle occupants are at increased risk of injury; this trend reaches statistical significance in the 7th decade (Moran et al, 2002; Peterson et al, 1998). The location of injury depends on which structures strike which car component and the severity depends on the speed and energy of the collision as well as timing of human contact to car structures. In a frontal collision an occupant continues to move forward as the vehicle stops. Forward motion of the occupant is arrested as the person connects either with the seatbelt or with anterior car structures, if unrestrained. Initial impact points are often lower extremities, resulting in fractures of the ankle, around the knee, or fracture of the femur. There are many factors that contribute to the amount of force transferred to specific anatomical structures including change in velocity at impact, timing of impact, degree of compartment intrusion, configuration of occupant and safety devices (Siegel et al, 2001; Bansal et al, 2009; Crandall et al, 1998; Nordhoff, 2004; Chong et al, 2007) Change in velocity at time of impact is closely associated with severity of injury as well as incidence of lower extremity injury (Figure 7) (Chong et al, 2007; Dischinger et al, 1998, Rupp & Scheider, 2004). The effect of timing is illustrated in the different degree of injury sustained when knee contact with instrument panel occurs during deceleration when the instrument panel may still be moving forward causing the localized contact velocity to be lower than impacts that occur once the car has stopped moving (Mackay, 1992). Occupant factors, such as age, gender, height and BMI also contribute to type and severity of injury. Height appears to be an important factor in pattern of injury; tall occupants (and males) sustain more knee, thigh or hip injuries while shorter (and female) occupants tend to sustain more foot and ankle injuries (Chong et al, 2007). Elderly occupants are at increased risk of injury (Moran et al, 2002; Peterson et al, 1998). There are studies that indicate that high BMI's are associated with increased severity of lower extremity injury (Arbabi et al, 2003; Boulanger e al, 1992).

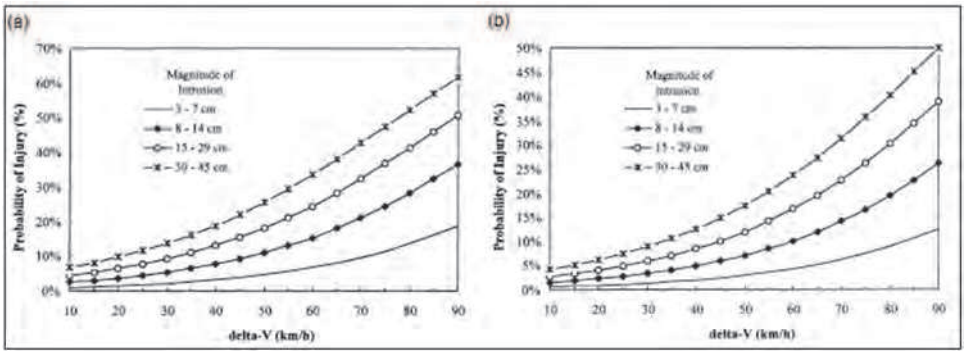

Fig. 7. Injury risk in a frontal MVC is related to magnitude of car intrusion and delta-V for both female (a) and male (b) occupants. Figure from Crandall et al (1998) with permission from Elsevier Limited

A typical dashboard injury pattern is often initiated by knee impact, usually on the instrument panel. This occurs most frequently in unrestrained occupants with or without airbag deployment. Force to the knee from the dashboard or instrument panel can result in knee laceration, patellar fracture, distal femur fracture, and proximal tibia fracture and forces can be transmitted through the femur to cause femoral shaft fractures, proximal femur fractures, acetabular or pelvic fractures, and posterior hip dislocation (Huelke, 1982; Rupp & Schneider, 2004). The risks for hip/pelvis injuries are generally greater than the risk for knee and thigh injuries at all crash severities and the right hip is more often injured than the left in forward-moving crashes, likely due to the effects of braking and bracing on occupant position and on muscle tension. Hip/pelvis fractures occur at lower impact force when the hip is flexed or adducted prior to impact; hip tolerance decreases approximately 1.8% for each degree of adduction from neutral and approximately 1% for each degree of flexion (Figure 8) (Rupp & Schneider, 2004). In an unrestrained driver, the body continues moving forward after the vehicle has stopped and the head, cervical spine and torso impact the windshield and steering wheel. During a lateral impact the occupant is accelerated away from the side of the vehicle that was struck and common injuries include lateral compression pelvic fracture, pulmonary contusion and intraabdominal solid organ injury (Mackay, 1992).

The other primary mechanism for lower extremity injury during a motor vehicle collision is impact caused by pedal interaction and toe pan intrusion (Crandall et al, 1998). One specific fracture that is well described is calcaneus or malleous fracture of the foot secondary to being forced against the brake pedal by the weight of the occupant or in combination with the floor pan of the car crushing into the space where the foot resides (Bucholz et al, 2005; Seipel et al, 2001). When the Achilles tendon resists dorsiflexion and the brake causes dorsiflexion, a three-point bending load occurs on the calcaneus with the posterior facet of the talus functioning as a fulcrum. This leads to a specific fracture pattern referred to as the tongue-toe calcaneous fracture pattern (Bucholz et al, 2005). Foot inversion or eversion in combination with compression force created by the brake pedal leads to malleolus fracture (Bucholz et al, 2005; Huelke, 1982; Crandall et al, 1998). High-heeled shoes have been shown to alter foot and ankle biomechanics leading to increased instability and injury during an MVC (Nordhoff, 2004). The effect of height on pattern of injury may be a reflection of leg position and may be related to the fact that shorter people sit closer to the steering wheel and reach for foot pedals, while taller people sit farther back with their knees closer to the level of the instrument panel (Chong et al, 2007).

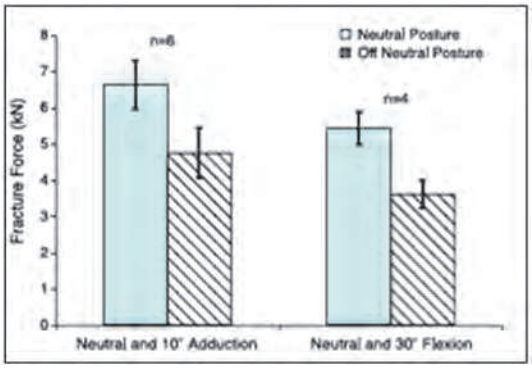

Fig. 8. Hip/pelvis fx occurs at lower impact forece when the hip is flexed or adducted prior to impact; Figure from Rupp & Schneider (2004) with permission from Elsevier Limited

Many safety systems, including seat belts, air bags, and vehicle deformation, take advantage of the fact that increasing time over which decelerations are applied to the passenger compartment decreases force experienced by the occupants (Peterson et al 1998). The concept of a "crumple zone" is based on this effect. Newer car designs provide an average of 2 ft of crushable car body, as well as steering mechanisms that collapse, which functions to increase deceleration time and also to dissipate a component of the energy by deformation (Peterson et al, 1998). Early goals of impact biomechanics and development of safety technology focused on decreasing mortality and head and thorax injuries to the extent that lower extremities are now the regions most likely to sustain injuries in frontal MVCs (Rupp & Schneider, 2004). These injuries are of substantial concern because they now account for up to 40% of treatment cost and nearly half of patients report significant long-term disability (Moran et al, 2002; MacKenzie et al, 2006).

Seat belts have had the single largest effect on reducing MVC-related mortality and injury, including extremity injury, decreasing fatalities by approximately 45-50% (Estrada et al, 2004; Cummins et al, 2008; Dihn-Zarr et al, 2001; McGwin et al, 2003). Seat belts increase deceleration time of the occupant via stretching of seat belt webbing and they distribute forces more uniformly on the body (Mackay, 1992; Peterson et al, 1998). There are multiple improvements in seatbelt technology that contribute to this effect. Pre-tensioners remove slack from the seatbelt upon detection of crash condition. Load limiters limits force imparted to the occupant by the seatbelt during the crash event by allowing the seatbelt webbing to yield when forces reach the set level. Web clamps lock the webbing to prevent or to minimize shoulder belt spool-out (Hinch et al, 2001).

Air bags are universally present in all new cars due to federal regulations, and it is well documented that they reduce risk of MVC-related mortality by 4-25% (Cummins et al, 2008; Dihn-Zarr et al, 2001; Estrada et al, 2004; McGwin et al, 2003). However, there is controversy regarding their effect on non-fatal injuries, particularly musculoskeletal injury. Air bag deployment without seat belt restraint is associated with increased incidence of lower extremity injury and some data suggests that air bag deployment together with seat belt restraint is also associated with increased incidence of lower extremity injury (Crandall et al, 1995; Cummins et al, 2008; Estrada et al, 2004; McGovern et al, 2000; Burgess et al, 1995; Chong et al, 2007). A possible contributing factor to the increased incidence of lower extremity injury is a "submarining" effect in which the pelvis and lower extremities are

shifted under the airbag and down the seat into the knee bolster and floor of the car (Estrada et al, 2004; Crandall et al, 1998; Cummins et al, 2008). Lower extremity trauma leads to significant impairments in function, and may be the most frequent cause of permanent disability after motor vehicle collision (MacKenzie et al 1993; Sieple et al, 2001). Improvement in safety systems directed at preventing lower extremity injury will be critical in the future. There is now increased interest in "knee bolster airbags" which could reduce the negative impact of airbags on lower extremity injury and in "smart air bags". These would be able to accurately sense the crash pulse, deploy in a graded fashion depending on the occupant size and weight, and deploy only when truly necessary (Peterson et al, 1998).

5.6 Anterior Cruciate Ligament (ACL) tear

Nearly 75% of ACL injuries are non-contact and occur as a result of self-initiated movement usually during athletic activities. The mechanism of injury is based on the anatomy of the knee. The ACL is a fibrous connective tissue that attaches the posterior aspect of the femur to the anterior aspect of the tibia. It courses anteriorly, medially and distally as it runs from femur to tibia. The primary function of the ACL is to limit anterior translation of the tibia relative to the femur.

ACL injuries are usually associated with decelerating and changing direction; often ACL injuries are caused by an internal twisting of the tibia relative to the femur or combination of torque and compression during a landing (Meyer & Haut, 2008). Despite intense study of the ACL injury during the past three decades, the exact mechanism of injury is not known (Boden et al, 2009). ACL injury occurs when an excessive tension force is applied to the ACL (Yu & Garrett, 2007). It has also been noted that in 96% of ACL tears, an opposing player is within close proximity, which could cause an alteration in the players' coordination leading to dangerous leg positions (Boden et al, 2009). There is some controversy as to how this force occurs, but based on recent studies, it is likely that an axial compressive force acting on the posterior tibial slope contributes to many ACL tears. This axial force results in posterior displacement of the femoral condyle on the tibial plateau, which applies tension to the ACL (Boden et al, 2010; Boden et al, 2009; Meyer & Haut, 2008). Boden et al found that subjects who experienced an ACL tear initially came into contact with the ground with their hindfoot or with their foot flat (the "provocative" landing position), whereas control subjects landed on the forefoot. It appears that normally, during landing, the foot, ankle, knee and hip joints work to dampen ground reaction forces. However, when subjects come into contact with the ground with the hindfoot or with their foot flat, the foot, ankle, and calf muscles are not able to absorb ground reaction forces, and the leg is converted into a two-segment column (above and below the knee), and the knee ends up absorbing a large component of the loading force. Additionally, under normal circumstances, as the calf muscles contract during absorption of ground-reaction forces, they produce a flexion force on the knee, activating the normal knee absorption mechanics. In the absence of calf muscle contraction, the knee may abduct or internally rotate rather than flex (Boden et al, 2009). Additionally, higher hip flexion angle at landing places the torso farther posterior to the knee, requiring that the quadriceps muscle must be activating during landing. The quadriceps muscle force provides anterior shear force on the proximal tibia which increases ACL strain (Boden et al, 2009; Yu & Garrett, 2007). Knee abduction (or knee valgus) also may play a role, particularly in female athletes, by potentially reducing the compression force threshold needed to produce ACL injury (Boden et al, 2010). However, it may also be that valgus collapse is the result of the ACL being torn rather than a cause (Boden et al, 2010; Meyer & Haut, 2008).

6. Experimental methods for injury dynamics

Our understanding of the dynamics of injury has progressed as a result of biomechanical studies, which have evolved from cadaveric, and laboratory studies in which mechanical injury is artificially produced, to finite element analysis where it is modeled quantitatively. Obvious ethical considerations have made difficult the study of actual injuries as they occur in real trauma situations, but techniques using video analysis of live injuries obtained from video clips posted freely by individuals (such as YouTube) are now becoming increasingly available. YouTube and other video-sharing sites provide a means to study the mechanics of injury in real live situations, on living subjects undergoing true physiological loading.

Mechanical in vitro tests on cadaveric structures have been performed for over a century, and the resulting data are the foundation for our understanding of the biomechanics of injury. Physiologic loading is an interaction between anatomical geometry, material properties of bone and soft tissue, and complex loading conditions. In mechanical testing, investigators are able to isolate loading parameters and to examine each systematically, as well as apply actual complex physiologic loads to a specific sample. The only advantage of using cadaveric subjects is the ability to subject the cadaver to impact loads and energies believed to be representative of those occurring in real-life trauma. Cadaveric specimens have been used to evaluate basic biomechanical parameters like strength, elastic modulus, toughness, anisotropy, how these properties change, and how bone reacts to various loading parameters. However, the impact responses of a cadaver specimen may significantly differ from a living human due to lack of muscular tone and neuromuscular reflexes, which can generate substantial deforming forces or protective tensioning during a traumatic event. Post-mortem changes in skin and fat and changes in the passive properties of muscles spanning the joints due to rigor or other embalming processes or freezing can also affect experimental results in unrealistic ways. Also, preservation methods may induce tissue damage that could alter results, and long investigation times can induce changes in mechanical properties of cadaveric bone over the course of the study, particularly if the bones are fresh, stored in a freezer, and repeatedly refrozen and re-thawed. (Zdero & Bougherara, 2010; Robinovitch et al, 1997)

Investigation of dynamics of injury performed on living humans has also contributed to understanding of the biomechanics of injury. Unlike cadaveric studies, forces due to muscle contraction and reflexes are taken into account, as well as physiological soft tissue properties. Additionally real-life fall scenarios with their inherent complexity can be investigated. However, studies using living humans are limited because they can only be performed at safe loading levels. The effects of higher loads must be extrapolated from the results obtained with lower loading parameters, and it is difficult to prove that the extrapolation is accurate, particularly with regards to biological tissue which displays non-linear force-deflection and force-velocity properties. Additionally, study subjects are often young while the actual event under study is most common in the elderly with bones and soft tissue that have different material and geometric properties. Extrapolating from young subjects to older subjects may be more unpredictable than extrapolating from low loads to higher loads. Furthermore, unlike cadavers, it can be difficult to confirm that subjects were performing the experiment as instructed and often experimental falls are self-initiated rather than random, likely representing "best-case" falls. (Robinovitch et al, 1997; Robinovitch et al, 2000; Chiu & Robinovitch, 1998; Choi & Robinovitch, 2010; Liang & Robinovitch, 2010)

Finite element analysis (FEA) is an engineering tool that has been used extensively in the study of the biomechanics of injury. It can be utilized to investigate intricate structures subject to complex loads, including those that occur in true traumatic injury, and it can estimate, with accuracy, how an object (a whole bone or trabecular network) behaves when subject to external loads. The object of interest is represented as a collection of a finite number of elements. The investigator must specify the boundary conditions (applied loads and/or applied displacements) and material properties. In this era FEA geometry can be obtained from CT or MRI scans which are then converted into three-dimensional geometries. Investigators can examine various stress and strain distributions, material properties, and energy densities and failure properties. Finite element analysis can provide estimates of quantities that are commonly obtained through mechanical testing (like whole bone stiffness), as well as quantities that are difficult, if not impossible, to measure experimentally (like strain density distributions). The behavior of bone at both the material and structural levels can be investigated. This technique has been particularly useful for understanding and predicting fracture risk, especially in complex situations. Finite element analyses can be performed under conditions that are difficult to create experimentally. However, how well the finite element solution approximates the actual biomechanical phenomenon depends critically on the quality of the data used as input. Uncertainty in choice of material properties and boundary conditions can severely limit the value of the results. Early FEA assumed two-dimensional geometries and used homogenous, isotropic elastic properties. Advances in computer hardware, CT, and model design now permit the development of more representative geometries and material properties. Quality of data will continue to improve with the use of high-resolution CT scans of anatomy and the use of non-linear material properties for human tissue and other advances. Other problems in FEA include challenges inherent in simulating objects that also undergo biologic processes, such as osteolysis, bone resorption, growth, or remodeling. Ultimately, investigators must be mindful of the fact that computer models are only as good as the information entered and that FE simulations should be validated by actual biomechanical experimentation when possible. (Mackay, 1992; Zdero & Bougherara, 2010)

The above experimental methods, despite their disadvantages, have provided us with important data that have played a critical role in helping us to understand the biomechanics of injury. However, although they can be used to provide accurate data, they cannot provide information from actual injuries. Ultimately these experimental methods are simulations and have to be interpreted as such. Cadaveric, laboratory, and computer simulation studies have been useful because studying the actual biomechanics of injury is limited by the obvious ethical and practical problems associated with conducting injury studies at physiological loading levels in live participants. However, video analysis provides an excellent opportunity to analyze the mechanism of injury in living subjects under physiological loading conditions. This technique is gaining interest, and its use will likely increase due to improved access to videos via video sharing sites, like YouTube. Kwon et al used ankle fractures obtained from posted YouTube videos to create a database of live ankle fractures occurring during diverse activities (Kwon et al, 2010). They used this methodology to evaluate the validity of the Lauge-Hansen ankle fracture classification system, a system developed using cadaveric models in the early 1950's to describe ankle fractures mechanistically. The Lauge-Hansen system has been recently challenged by more modern cadaveric biomechanical study (Michelson) and an MRI imaging study (Gardner) revealing that the sequences of injures predicted from cadavers were not actually reproducible with

modern techniques or visualized in the sequence described when MRIs were studied from true ankle fractures. YouTube searches generated videos of ankle trauma, and the individual posting the video was then offered participation in the study. Inclusion criteria required that the video demonstrated clear visualization of the mechanism of injury, that the subject sustained a fracture or dislocation, and that x-rays also revealed ankle fracture. Videos and radiographs were independently analyzed and categorized by mechanism of injury according the Launge-Hansen classification system. The radiographs and videos were then examined for consistency. The case series suggests that Launge-Hansens's mechanistic classification of radiographs does not correlate to the actual injury mechanism; the Lauge-Hansen system was only 58% overall accurate in predicting fracture patterns from the deforming injury mechanism. The classification system performed particularly poorly at predicting pronation external rotation type fracture patterns.

Other studies using video analysis to evaluate injury mechanism used athletic game footage. Giza et al evaluated game footage to determine the mechanism and weight-bearing status that placed soccer players at risk for foot and ankle injury (Giza et al, 2003). Kroshaug et al and Boden et al analyzed videos of athletic events to study the mechanism of ACL injury (Kroshaug et al, 2006; Boden et al, 2009). Andersen et al analyzed videos of game footage to describe injury mechanism for ankle injury in elite male football (soccer) (Andersen et al, 2004). There are disadvantages of this experimental methodology, which must be taken into account. Videos are collected as convenience samples, and therefore the camera angle is not always ideal for analysis, and clothing can make identifying anatomic landmarks difficult. Furthermore, it can be difficult to determine the exact moment at which the injury occurred; abnormal movements occurring after the injury could be confused for the mechanism of injury, as is clear from the controversy about abduction as either a cause or the result of ACL tear. The largest hurdle experienced by Kwon et al was their difficulty recruiting subjects from the internet-based video sharing site, YouTube.

7. Conclusion

Traumatic injuries in general, and fractures in particular, represent an important health concern and affect the majority of people at some point in their lives. There is an array of study techniques that take different approaches to studying the biomechanics of the musculoskeletal system, which have provided a basic understanding of bone properties and have helped to explore how they change with age. This has facilitated investigation of the mechanics of fracture and injury since the mechanical properties of the musculoskeletal system determine when and how structures will fail. In particular, the ability to study injury while it occurs under true physiologic loading conditions via video-sharing websites is an important tool that will likely continue to provide new insights into fracture mechanism. Strides have been made, based on this information, to develop means of decreasing or preventing injury. While a great deal is known about the biomechanics of injury and fracture healing, challenges remain, and areas of future study will be proposed.

8. References

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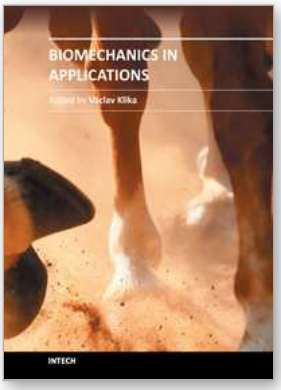

Biomechanics in Applications

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During last couple of years there has been an increasing recognition that problems arising in biology or related to medicine really need a multidisciplinary approach. For this reason some special branches of both applied theoretical physics and mathematics have recently emerged such as biomechanics, mechanobiology, mathematical biology, biothermodynamics. The Biomechanics in Application is focusing on experimental praxis and clinical findings. The first section is devoted to Injury and clinical biomechanics including overview of the biomechanics of musculoskeletal injury, distraction osteogenesis in mandible, or consequences of drilling. The next section is on Spine biomechanics with biomechanical models for upper limb after spinal cord injury and an animal model looking at changes occurring as a consequence of spinal cord injury. Section Musculoskeletal Biomechanics includes the chapter which is devoted to dynamical stability of lumbo-pelvi-femoral complex which involves analysis of relationship among appropriate anatomical structures in this region. The fourth section is on Human and Animal Biomechanics with contributions from foot biomechanics and chewing rhythms in mammals, or adaptations of bats. The last section, Sport Biomechanics, is discussing various measurement techniques for assessment and analysis of movement and two applications in swimming.

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InTech Europe
University Campus STeP Ri
Slavka Krautzeka 83/A
51000 Rijeka, Croatia
Phone: +385 (51) 770 447
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Unit 405, Office Block, Hotel Equatorial Shanghai
No.65, Yan An Road (West), Shanghai, 200040, China
中国上海市延安西路65号上海国际贵都大饭店办公楼405单元
Phone: +86-21-62489820
Fax: +86-21-62489821