

Understanding Bone Strength from Finite Element Models: Concepts for Non-engineers

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Abstract Recent clinical studies have reported not only changes in bone mineral density in patient populations but also changes in bone strength determined using finite element modeling. Finite element modeling is a technique well established in engineering but unfamiliar to many clinicians and basic biologists. Here, we provide a conceptual introduction to finite element modeling and its clinical applications to bone that is written for individuals without any background in engineering. Finite element modeling of bone is the net result of over 60 years of effort in the engineering community and over 40 years of effort in the field of bone biomechanics. We discuss the mathematical and theoretical basis for finite element modeling, how finite element models are created from clinical images and the assumptions made in using finite element models to estimate whole bone strength. In addition, we discuss the limitations of finite element modeling in patient populations with altered bone tissue quality. Clinical studies have shown that prediction of fracture risk using finite element modeling is as effective, and in some cases superior, to simple measures of bone mineral density. Further application of finite element modeling to clinical studies has the potential to improve fracture risk assessment beyond what is currently possible with bone mineral density.

Keywords Osteoporosis · Bone quality · Bone strength · Biomechanics · Finite element

Introduction

Osteoporosis is characterized by impaired bone density and bone strength, leading to an increased risk of fracture from normal daily activities. The primary means of diagnosing osteoporosis is assessment of bone mineral density using dual-energy X-ray absorptiometry (DXA). While bone mineral density is a useful screening tool, many osteoporosis-related fractures occur in patients who are not identified as high risk based on bone mineral density, indicating a need for methods to further improve fracture risk prediction.

In 1994, the World Health Organization developed a criterion for diagnosis of osteoporosis based on bone mineral density [1]. A patient with bone mineral density values more than 2.5 standard deviations below the average seen in a young adult female population was considered to have osteoporosis while patients with bone mineral density levels between 1.0 and 2.5 standard deviations below the mean are diagnosed with osteopenia and considered at risk of development of osteoporosis. Additional assessment of fracture risk can be performed by accounting for patient factors (sex, ethnicity, smoking status, etc.) using a tool such as FRAX [2, 3]. While statistical tools such as FRAX are useful for assessing fracture risk, they are limited in that they require sufficient clinical data on fracture risk associated with an influential parameter (age, race, etc.). In patient populations without sufficient clinical fracture data, our confidence in estimates of risk based on statistical methods is reduced.

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A major limitation of bone mineral density and FRAX is that they do not directly account for the physical mechanisms that lead to fracture. Bone mineral density can describe the shape and density of a bone, but does not directly evaluate how well a bone's internal structure and material properties resist physical forces. The ability of structures and materials to resist physical forces, however, has long been studied by engineers. One of the most powerful modern engineering tools for assessing the ability of a structure or material to resist mechanical failure is finite element modeling. Finite element modeling is a computer simulation approach that is used to determine how applied loads are distributed within a structure and how well that structure resists loads. Over the last 20 years, finite element modeling of bone has transitioned from a tool for basic and translational science into a method used in clinical studies (for thorough reviews see [4–6]). The application of finite element models to clinical studies has placed many clinicians and researchers without engineering training in the uncomfortable situation where they must evaluate clinical results derived from finite element models and perhaps apply that knowledge to patient care.

The purpose of the current review is to explain the fundamentals of finite element modeling to clinicians and others without backgrounds in engineering. The article is, in part, a response to conversations the senior author has had with clinicians and biologists who see finite element modeling as a potentially useful approach to studying bone but have difficulty weighing the importance of data from finite element models relative to other findings. To help individuals without an engineering background understand the basis for finite element models, the current manuscript assumes no prior knowledge of mechanics beyond introductory college physics and conveys concepts without the use of mathematical equations. We have also limited the discussion to finite element modeling as applied to clinical images and do not discuss more recent advancements in finite element modeling of bone that have only been used in preclinical studies. We have, necessarily, glossed over many subtle and not-so-subtle details of the finite element method and refer the curious reader to more technical reports for more information.

How Materials Resist Mechanical Loads: Stiffness, Strength, Stress, Strain

To understand how a finite element model works, it is necessary to first discuss how materials respond to physical forces. Consider a rubber band. When a force is applied to the rubber band, it stretches and if an even larger load is applied it stretches further. The relationship between the amount of applied load and the resulting deformation is the stiffness of the object and is described in introductory

physics textbooks using a spring constant. There are limits to the amount of load/deformation that can be applied to an object. The maximum load that an object can carry before mechanical failure is the object's strength. The stiffness and strength of an object are determined by object size and shape as well as the properties of the material from which the object is made. The rubber band we have described could be made stiffer and stronger either by changing its shape (making it thicker) or by making it from a stiffer and stronger form of rubber. To separate the contributions of object size and material properties, we normalize stiffness and maximum load by the size. Load normalized by the cross-sectional area of the object is referred to as stress, and deformation normalized by the original length of the object is known as strain. The relationship between stress and strain is the stiffness of the material (as opposed to the stiffness of the entire object), referred to in the engineering literature as the Young's modulus.

Deformations from small loads are reversible, that is, when the load is removed the material recovers its initial shape. Deformations that recover after the load is removed are known as elastic deformations. All materials display elastic deformation under small loads. When applied stresses and strains are too large, however, permanent deformations are generated. Permanent deflections can be generated even when the material has not ruptured. For this reason, the strength of a material can be expressed in two different ways: (1) the smallest stress at which the material begins to deform permanently and (2) the maximum stress that the material can sustain. A paperclip is a common object in which permanent deflections are applied without rupture. It is possible to apply loads to a paperclip to permanently change its new shape, but larger stresses are required to break the paperclip in two.

We have used the examples of the rubber band and paperclip because these are common objects that are compliant enough for their deformations to be observed with the naked eye. All materials, including bone, undergo deformation when loaded, although in materials with very high stiffness the deformation is microscopic, while we have so far discussed only tensile loads (those that involve stretching the object). Mechanical stresses are also generated in objects submitted to compression (compressive stresses) or torsion/twisting (shear stresses). The stiffness and strength of bone as a material have been studied under many different loading scenarios providing sufficient understanding of the material properties of bone to generate useful finite element models.

What is a Finite Element Model?

Finite element modeling is an engineering technique developed over the last 60 years to understand how stresses

are distributed in a complicated structure. Finite element modeling is a standard technique used by engineers to ensure that a machine designed to transmit forces will not fail when put into service. Many bachelor's degree programs in mechanical engineering require at least an introduction to finite element modeling. In the aerospace industry, finite element modeling has been used as the primary tool for aircraft design for over 20 years. Aircraft manufacturers are so confident in the accuracy of finite element models that they no longer perform many of the mechanical tests that were once considered necessary when designing a new vehicle. The adoption of finite element modeling has therefore allowed aircraft manufacturers to save considerable time and effort without sacrificing safety. The success of finite element modeling in the aerospace industry illustrates the potential for the approach to reliably estimate bone strength and potentially fracture risk.

At the most fundamental level, finite element modeling is a method of solving differential equations using numerical methods on a computer [7]. Differential equations are mathematical relationships in which the rate of change of a parameter is known and it is desirable to calculate the actual value of the parameter at a later point in time. The differential equations describing most finite element models have not yet been solved analytically and therefore require the use of numerical methods on a computer. A numerical method familiar to most readers is the trapezoidal rule, which is a means of determining the area under a curve. When using the trapezoidal rule, one divides the region under the curve into discrete segments and fits a trapezoid to the curve within each region. The sum of the areas of all of the trapezoids provides a useful estimate of the area under the curve. Just as the trapezoidal rule requires dividing up the problem into discrete segments, the finite element method divides a three-dimensional object into small regions called elements. By describing the relationship between stress and strain within each element, it is possible to calculate the distribution of stress in the object when a load is applied. Readers familiar with the trapezoidal rule may remember that the approach is sensitive to the initial conditions and the size of the discrete segments (smaller segments are more accurate). Finite element models are also sensitive to initial conditions, which commonly include applied loads and boundary conditions. The size of the individual elements in a finite element model also influences results, with smaller element size providing improved resolution.

Creating Patient-Specific Finite Element Models to Evaluate Bone Strength

The primary clinical application of finite element models in bone is estimation of whole bone strength. Here, we

conceptually describe the steps in creating patient-specific finite element models. Readers interested in more technical details are referred to other sources [4, 8, 9].

Patient-specific finite element models are generated from clinical images. The most commonly used patient-specific finite element models are created from “quantitative computed tomography” or QCT. Quantitative computed tomography differs from standard computed tomography in that a mineral density calibration phantom is scanned along with each patient. A mineral density calibration phantom is a plastic plate containing a series of tubes of fluid with known concentrations of mineral. By scanning the calibration phantom at the same time as the patient, it is possible to relate the brightness of pixels and voxels within the CT image (measured in Hounsfield units) to mineral density values. The calibration phantom is required because scan parameters and X-ray tube output can vary from scan to scan (even with the same scanning device) and must be adjusted using the calibration curve to achieve sufficient accuracy in bone density.

Once an image is acquired, a series of image processing steps are used to generate the finite element model itself. These steps include isolating the bone from soft material and dividing the image into finite elements. Isolating the bone from the soft tissue is typically performed using an image processing technique called thresholding or segmentation (please see [8, 10] for more technical details). Once the bone is isolated from surrounding soft tissues, it consists of a three-dimensional image made of thousands to millions of voxels (a voxel is a three-dimensional pixels). The brightness of each voxel in the image corresponds to the density of bone tissue at that location. Density is determined at each voxel based on the relationship between brightness and mineral density determined using the calibration phantom. The local bone tissue density is then related to bone tissue Young's modulus using empirical relationships taken from bone biomechanical studies [11, 12]. The result is a three-dimensional model of the bone in which the internal density distribution is represented as a distribution of bone tissue stiffness (Fig. 1).

Once the shape and density distribution of the bone are determined, the finite element model can be implemented and used to estimate bone strength. Implementation involves selecting applied loads and displacements that simulate a particular loading situation. Commonly loads applied in finite element models of bone mimic either habitual loading (standing, walking, etc.) or an overload such as a fall from standing height. An example of a habitual loading situation is compression of the vertebral body, which is simulated by applying displacements to the cranial side of the vertebral body while constraining displacement at the caudal side (Fig. 1). Finite element models of the proximal femur used with clinical images

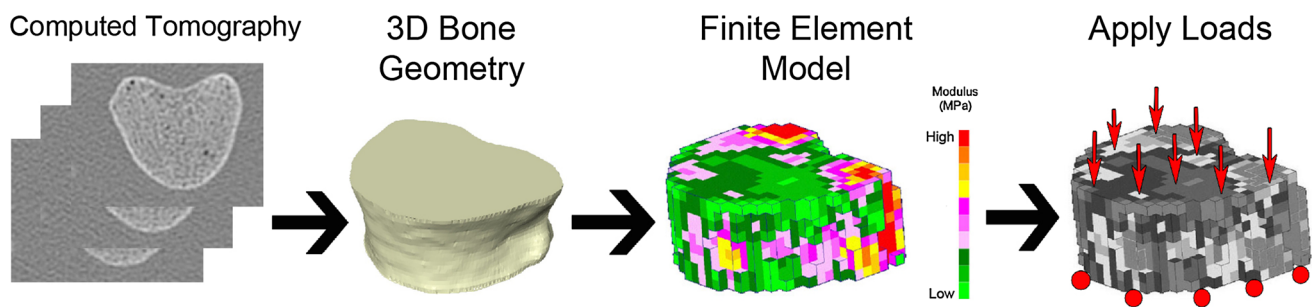


Fig. 1 Steps in creating a finite element model are illustrated. The quantitative computed tomography image of the patient vertebral body is converted into a three-dimensional image. The distribution of bone mineral density is shown by variation in *color* among image

voxels. Each voxel is converted into an element, and the finite element simulation is implemented by applied loads or displacements (*arrows*) while constraining some surfaces (*circles*) (Color figure online)

mimic loads during standing, or alternatively, loads associated with a fall to the side (Fig. 2). Selecting the appropriate loading situation and associated loads and boundary conditions requires engineering expertise.

The results of the finite element analysis include the stress and strain within each element of the model, which can consist of thousands to millions of numerical values. Relating such a large set of information to clinical fracture requires a failure criterion. The most simple failure criterion would be to define whole bone failure as the condition when any portion of the bone tissue experiences stresses in excess of tissue strength. However, failure of one very small region of the whole bone is unlikely to result in clinical fracture and often a more complicated failure criterion must be defined. Some failure criteria used successfully to predict whole bone failure include the applied load that results in excessive deformation of the entire bone or failure of an excessive proportion of the tissue within the whole bone. Failure criteria are commonly validated using biomechanical testing of whole bones in the laboratory [12–14].

The computer resources required to implement a finite element model represent a balance between model resolution and time. A relatively simple finite element model of a whole bone at low resolution (each element ~ 1 mm in characteristic size) can require minutes to hours of computational time on a standard desktop computer, while some of the higher resolution models (each element ~ 10 μ m in characteristic size) can take more than 4 h of operating time using a national supercomputer resource. More complicated finite element models (for example, those that include nonlinearities in bone mechanical performance) have computational demands more than ten times greater than the more simple models.

Interpreting Finite Element Models of Whole Bones

Finite element modeling itself, as we have mentioned previously, is a well-validated approach for the design of

engineering components. There are many reports, demonstrating that finite element model-derived strength can predict fracture risk in the spine and hip, in many cases in a manner that is independent of bone mineral density (see [9] for a recent review). While evaluating the technical details of finite element methodologies requires engineering expertise, there are three primary assumptions of all currently used finite element modeling approaches that are relevant to interpretation in the clinic.

First, finite element models describe mechanical performance of a bone in just one loading situation (cranial–caudal compression in a vertebral body, a fall to the side or loads from standing on the proximal femur, etc.). These loading situations are selected to mimic situations relevant to whole bone failure. However, each loading condition will provide a distinct assay of bone strength (whole bone strength depends on how the loads are applied). The estimation of whole bone strength is therefore most accurate in the specific loading situation simulated and can also be considered representative of fractures occurring in other loading situations. However, finite element models may be less effective when applied to patient populations in which fragility fracture is considerably different from the simulated loads.

Second, the finite element models we have described evaluate failure using bone tissue strength, which describes failure of the bone due to a discrete overload. Not all clinical fractures are caused by an isolated overload. Stress fractures, for example, occur as a result of damage accumulation from hundreds of cycles of loading [15]. Finite element simulations that include damage accumulation and propagation are complicated mathematically and have not yet been applied to clinical image-based finite element models. Hence, current finite element modeling approaches may not be as effective in predicting fractures resulting from more than one overload. Recently, there has been increased interest in bone tissue ductility or bone tissue toughness as a contributor to fracture risk [16]. Bone tissue ductility and toughness are material properties that describe

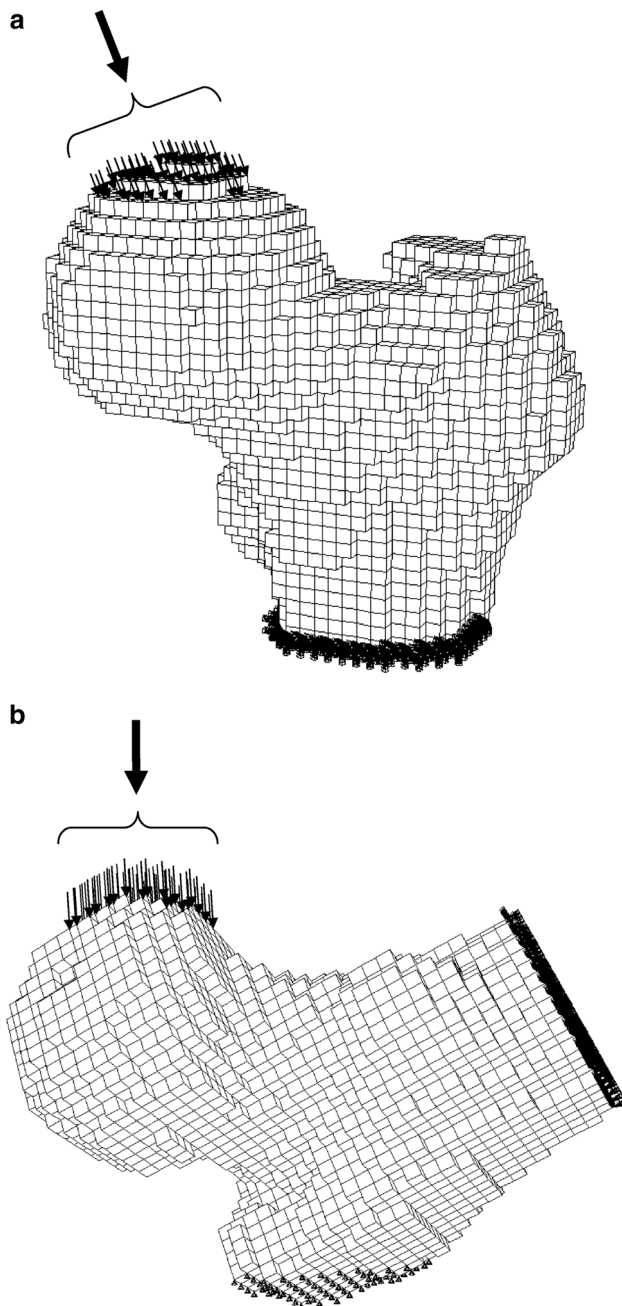


Fig. 2 Finite element models of whole bones may simulate different loading situations. **a** Loads and boundary conditions to simulation loads on the proximal femur during standing/walking. **b** Loads and boundary conditions used to simulate a fall to the side. Reprinted from Keyak et al. [24] with permission from Elsevier

the ability of the bone tissue to absorb energy; bone tissue with low ductility tends to fail more like glass, while bone tissue with high ductility can deform considerably before failure (consider a green-stick fracture). There are relatively few finite element modeling approaches that include the effects of tissue ductility and those that are currently available require considerably more computational

resources and have therefore seen little use in whole bone finite element models to date. As a result, the current finite element models are expected to be less effective at predicting fracture in patient populations in which bone tissue ductility is altered.

Third, as we have mentioned above, finite element models of whole bones are generated by assigning a Young's modulus to each voxel of the clinical image. The assignment is performed by using empirical relationships between bone tissue density and bone tissue stiffness and strength. While these relationships are quite useful, they are generated from best fit curves that display some variability. Additionally, the empirical relationships are based on biomechanical studies of bone tissue from otherwise health donors. Such relationships may not be representative of all patients, especially those with disorders that affect bone and mineral metabolism. For example, in some patient populations fracture risk exceeds what is expected from measures of bone mineral density, suggesting that the relationship between tissue strength may be less than expected from density [17]. Large changes in the relationship between tissue mechanical properties and tissue density will make predictions of whole bone strength from finite element models less accurate. There is evidence that the relationship between tissue density and stiffness and strength is altered in patients with a history of glucocorticoid treatment [18] and patients with diabetes [19] and it is possible that finite element models applied to those patient populations may not be as effective as they are in normal patients.

Despite these limitations, finite element modeling remains a well-validated approach for estimating whole bone strength from clinical images [4–6, 9]. Results from patient-specific finite element models are correlated with clinical fracture [20] and can also be used to noninvasively monitor improvements in bone strength associated with interventions including pharmacological treatment [21–23].

Conclusions

Finite element modeling is a computational approach based on over 60 years of mathematical and computational development and experimental validation by the engineering community. The application of finite element modeling to clinical images of bone represents clinical translation of over 40 years of basic research into the mechanical properties of bone. While some improvements in fracture risk prediction are still to be made, finite element modeling is an extremely useful tool for assessing bone strength in patients.

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Compliance with Ethical Standards

Conflict of interest Christopher J. Hernandez and Erin N. Cresswell have no potential conflict of interest.

Animal and Human Studies This article does not include any studies with human or animal subjects performed by the author.

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