

Article

Comparative Experimental and Viscoelastic Modeling Study of Human Tibial Trabecular Bone Under Healthy and Osteoarthritic Conditions

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Abstract

Background: The development of osteoarthritis (OA), a whole-joint disorder that is increasingly recognized, depends on subchondral trabecular bone. Variations in the viscoelastic characteristics of trabecular bone have been proposed to affect load distribution and potentially lead to joint degradation. The viscoelastic response of human trabecular bone in both healthy and osteoarthritic situations was investigated using constitutive modeling and stress-relaxation testing. Fifteen tibial trabecular bone specimens were evaluated using Standard Linear Solid (SLS) models and two-branch generalized Maxwell models following uniaxial stress-relaxation testing. Mechanical, energy, and relaxation-related traits were retrieved and compared between groups. **Results:** Healthy bone tended to relax stress more slowly and to bear mechanical loads over time to a slightly greater extent than osteoarthritic bone, which tended to relax stress more quickly and had poorer mechanical endurance; these differences were not statistically significant. The Generalized Maxwell model suited the experimental data better than the SLS model ($R^2 > 0.98$), capturing both short- and long-term relaxation mechanisms. Sensitivity analysis revealed higher parameter variability in OA specimens, suggesting possible differences in mechanical heterogeneity and load-dissipation behavior that require further investigation. **Conclusions:** Although the observed differences were not statistically significant in this exploratory study, the results suggest potential trends toward altered viscoelastic behavior between healthy and osteoarthritic trabecular bone. Future studies with larger cohorts are needed to further investigate osteoarthritis-related biomechanical alterations. Multi-branch viscoelastic modeling may provide sensitive mechanical descriptors for characterizing the relaxation behavior of subchondral bone.

Keywords: osteoarthritis; trabecular bone; viscoelasticity; stress-relaxation; generalized maxwell model; biomechanics; standard linear solid (SLS)



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1. Introduction

Osteoarthritis (OA) is one of the most prevalent chronic musculoskeletal disorders and a major cause of pain, disability, and reduced quality of life worldwide. This condition leads to disability and a poorer quality of life worldwide. The global burden of OA has risen substantially in recent decades and is projected to rise due to population aging,

increasing obesity rates, and metabolic risk factors [1]. Moreover, moderate-to-severe OA is associated with substantial functional limitations, chronic pain, and a significant impact on individuals' quality of life, in addition to its high prevalence in healthcare systems and society [2]. Recent epidemiological analyses and international reports have underlined the urgent need to improve knowledge on OA pathophysiology and to develop more effective strategies for prevention and treatment [1,3,4].

In the past, OA was mainly regarded as a cartilage degenerative disease, but recent evidence supports the concept of OA as a whole joint disease affecting multiple tissues, including articular cartilage, synovium, ligaments, and subchondral bone [5,6].

The subchondral bone is important in the mechanical functioning of the joint, where it supports the transfer of load from the cartilage to the underlying skeletal structure. Hu et al. [5] have shown that the microenvironment changes of subchondral bone, including abnormal remodeling activity, vascular changes, and inflammatory responses, actively contribute to OA progression and pain development. Similarly, Fan et al. [6] emphasized the importance of changes in the osteochondral interface in the formation and progression of cartilage degeneration. However, these studies were mainly centered on biological and structural mechanisms, and the direct relationship between these alterations and the mechanical behavior of trabecular bone is still not sufficiently understood.

The mechanical properties of trabecular bone are largely determined by its hierarchical architecture, including trabecular thickness, connectivity, bone volume fraction, mineralization, and marrow composition. Oftadeh et al. [7] thoroughly reviewed that the trabecular organization controls load transmission, stiffness, and energy dissipation in multiple structural scales. More recently, imaging-based studies have confirmed significant microstructural changes in the subchondral bone induced by OA, such as sclerosis, modified trabecular organization, and changes in remodeling patterns of the bone [8,9].

These investigations significantly improved the understanding of structural deterioration during OA. However, most imaging studies remain descriptive and provide limited information regarding how these microstructural changes affect the time-dependent mechanical response of trabecular bone.

Progressive development of constitutive modeling approaches has been undertaken to better describe bone mechanical behavior. Pahr and Zysset [10] reviewed the recent advances in constitutive modeling of bone and showed the importance of incorporating nonlinear elastic, viscoelastic, and multiscale formulations to accurately predict the bone deformation under physiological loading conditions. Micheletti and Shah [11] also demonstrated that the mechanical behavior of bone is a result of its hierarchical organization, where structural and compositional features at different length scales govern the overall mechanical response. These advances in multiscale biomechanics provide important insights into bone behavior; however, many existing computational models remain insufficiently validated against experimental data, particularly from human osteoarthritic bone.

Experimental studies have shown that biological tissues associated with OA have complex time-dependent mechanical behavior arising from interactions between solid matrix deformation and fluid redistribution. In early studies by Mow et al. [12], creep and stress-relaxation behaviors were first shown to be fundamentally due to fluid flow in porous biological tissues, and the biphasic theory of cartilage mechanics was first developed. More recently, Ebrahimi et al. [13] studied the dynamic viscoelastic properties of osteoarthritic cartilage and showed that the tissue degeneration associated with OA alters time-dependent mechanical behavior. These studies have been important in elucidating the mechanical consequences of OA in cartilage but have focused primarily on articular cartilage. The viscoelastic response of osteoarthritic trabecular bone is poorly characterized.

The theoretical underpinnings of creep and stress relaxation offered by Findley et al. [14] can be used to understand nonlinear viscoelastic behavior in the context of bone mechanics. In a recent work, Wu et al. [15] developed and validated a viscoelastic constitutive model for human cancellous bone and demonstrated that generalized viscoelastic models may accurately reproduce experimental relaxation responses. However, rather than directly comparing real trabecular bone from healthy and osteoarthritic individuals, their work focused on model building and validation. Therefore, the effect of OA on trabecular bone relaxation kinetics, energy dissipation, and viscoelastic parameters is not well known. Fluid transport through the porous trabecular network also contributes to the time-dependent mechanical response of bone via poroelastic mechanisms. In addition, recent studies on biological tissues have shown that extracellular matrix properties and fluid redistribution mechanisms strongly influence viscoelastic relaxation behavior over multiple time scales [16]. However, the contribution of altered fluid–solid interactions to the mechanical deterioration of osteoarthritic trabecular bone remains poorly quantified, despite these theoretical advances.

The recent development in imaging techniques and computational modeling has improved the ability to characterize the structural changes associated with OA. Khokhar and Conaghan [17] reviewed the role of advanced imaging approaches in evaluating osteoarthritis progression and highlighted the ability of imaging biomarkers to detect subchondral bone alterations. However, imaging techniques alone cannot fully characterize the functional mechanical consequences of these structural changes.

In a similar vein, multiscale modeling studies such as that of Xiong et al. [18] have shown the importance of incorporating structural information into computational predictions of bone behavior. But a critical limitation remains without direct experimental validation using human osteoarthritic trabecular bone and a quantitative viscoelastic analysis.

While important advances have been made in understanding bone biomechanics, many critical questions still have not been addressed. In particular, few studies have combined experimental stress-relaxation measurements, constitutive viscoelastic modeling, and energy-based mechanical descriptors concurrently to compare healthy and osteoarthritic human trabecular bone. In addition, the link between OA-related microstructural degradation and alterations in relaxation kinetics, dissipation capacity, and model parameters is still not well understood. Addressing this knowledge gap is essential for improving the biomechanical understanding of subchondral bone dysfunction and for developing more accurate computational models of OA progression.

Therefore, the present study aims to provide a comprehensive experimental and numerical comparison of human tibial trabecular bone obtained from healthy and osteoarthritic donors. Stress-relaxation tests were performed under constant strain conditions ($\epsilon_0 = 0.1667$; displacement limit = 1.5 mm), and time-dependent mechanical parameters, including initial stress (σ_0), equilibrium stress (σ_∞), half-relaxation time ($t_{1/2}$), relaxation ratio, energy components, and stress–time integral, were quantified. Experimental relaxation responses were analyzed using two complementary constitutive approaches: the Standard Linear Solid (SLS) model and a two-branch generalized Maxwell model [10,16]. Sensitivity analyses were additionally performed to identify the parameters governing early and late relaxation behavior and to evaluate model robustness. This integrated experimental–numerical framework provides new insights into the viscoelastic behavior of human trabecular bone and contributes to a better understanding of the mechanical consequences of osteoarthritic remodeling.

The integration of experimental stress-relaxation testing, energy analysis, multiscale constitutive modeling, and sensitivity analysis in human osteoarthritic trabecular bone is still limited. Prior research has examined bone viscoelasticity using experimental or com-

putational methods. Thus, the goal of this research is to present a thorough biomechanical description of human tibial trabecular bone in both healthy and OA situations. The connection between time-dependent mechanical behavior and osteoarthritis-related alterations in human trabecular bone is still unclear despite prior research on bone viscoelasticity, especially when using combined experimental and constitutive modeling techniques.

2. Materials and Methods

2.1. Ethical Approval

This study was approved by the Ethics Committee of La Rabta University Hospital (Tunis, Tunisia) and conducted in accordance with the Declaration of Helsinki. Written informed consent was obtained from all participants prior to tissue collection. Human trabecular bone specimens were collected from adult patients undergoing knee arthroplasty surgery.

2.2. Sample Collection and Preparation

Fifteen human trabecular bone specimens were collected from tibial epiphyses during orthopedic surgical procedures performed at Rabta Hospital (Tunis, Tunisia). Osteoarthritic specimens were obtained from patients undergoing total knee arthroplasty for advanced knee osteoarthritis, whereas healthy specimens were collected during orthopedic surgery for traumatic conditions without clinical or radiographic evidence of osteoarthritis or other degenerative joint diseases. Group allocation was established according to the clinical diagnosis, medical history, preoperative imaging findings, and dual-energy X-ray absorptiometry (DEXA) assessment. Specimens were collected from eight donors (6 males and 2 females; age range: 19–64 years). Donors were classified as either healthy (Healthy, $n = 8$) or osteoarthritic (OA, $n = 7$). After removal of the articular cartilage, samples of subchondral trabecular bone were carefully removed using a diamond saw and machined into cubic specimens with dimensions of $20 \times 20 \times 20 \text{ mm}^3$. The principal trabecular orientation was preserved during sample preparation, and specimens were positioned to maintain the predominant anatomical trabecular alignment during mechanical testing. Specimen dimensions were verified using a digital caliper before testing. During specimen preparation and testing, samples were handled carefully to minimize dehydration and preserve their native hydration state. Although no prolonged storage was applied, variations in tissue hydration cannot be completely excluded and may influence the measured viscoelastic response.

2.3. Stress-Relaxation Testing

A universal testing apparatus (LLOYD EZ50, Lloyd Instruments, Bognor Regis, UK) with a 5 kN load cell was used to conduct mechanical tests. At room temperature ($22 \pm 1 \text{ }^\circ\text{C}$), specimens were tested in unconfined compression between two parallel aluminum platens (diameter: $50 \pm 0.1 \text{ mm}$).

Prior to testing, a preload of 5.6 N was applied to ensure complete contact between the specimen and the compression platens (Figure 1). Compression was applied along the principal trabecular orientation at a displacement rate of 0.5 mm/min until a displacement of 1.5 mm was reached, corresponding to an imposed strain of $\epsilon_0 = 0.1667$. It was chosen to produce a quantifiable stress-relaxation response while being inside the typical deformation range for trabecular bone specimen mechanical characterization. Instead of directly simulating physiological strains that occur during typical activities, this strain level was selected for comparative mechanical testing under controlled experimental settings. After that, the strain was kept constant for about 1800 s while the stress response was continuously monitored.

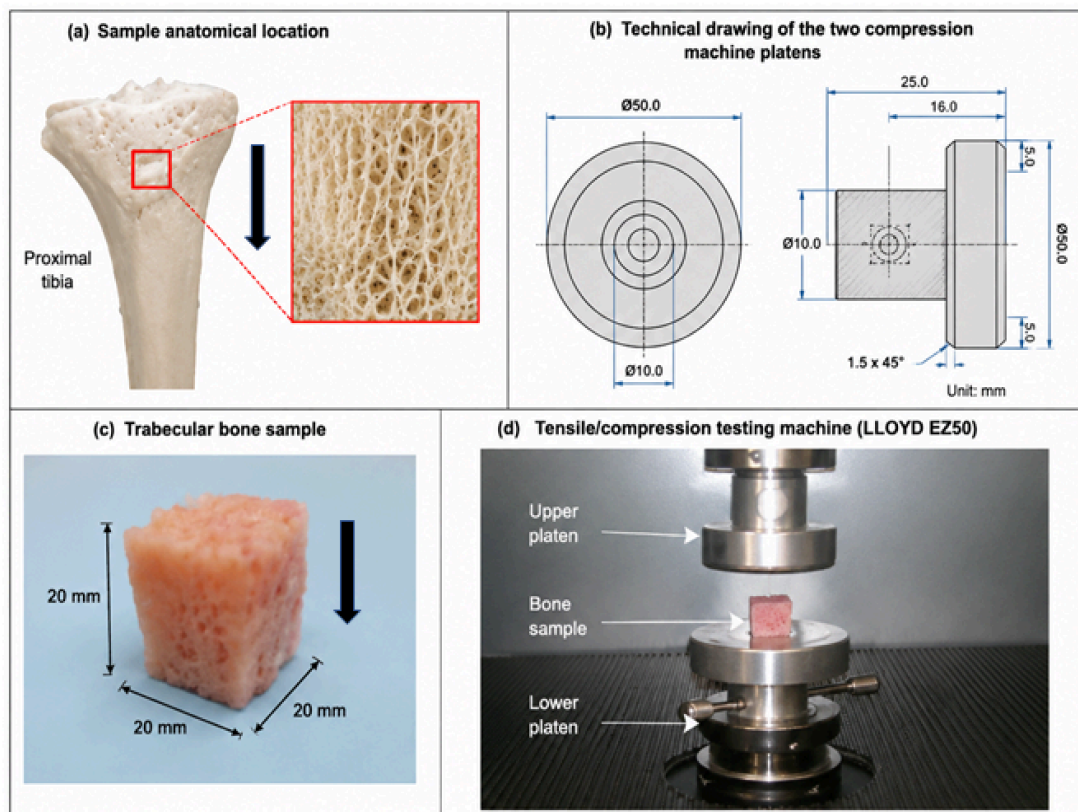


Figure 1. Compression/Relaxation test: (a): Sample anatomical location, (b): Technical drawing of the two compression machine platens, (c): trabecular bone sample (a cube with 20 mm sides), (d): Tensile/compression testing machine (LLOYD EZ50) (two special small-size cylindrical compressing trays (diameter 50 mm $\pm$ 0.1)).

It should be mentioned that rather than using an ideal instantaneous step strain, a finite loading ramp was used to achieve the applied strain. The viscoelastic parameters discovered in this study should thus be regarded as effective relaxation parameters under the specified experimental conditions. Meaningful comparisons between the OA and healthy groups are still valid because all specimens were tested according to the same protocol.

2.4. Rheological Models

The experimental stress-relaxation data were fitted using two viscoelastic constitutive models: the SLS model and a generalized Maxwell model composed of two Maxwell branches connected in parallel with an elastic spring (Figure 2).

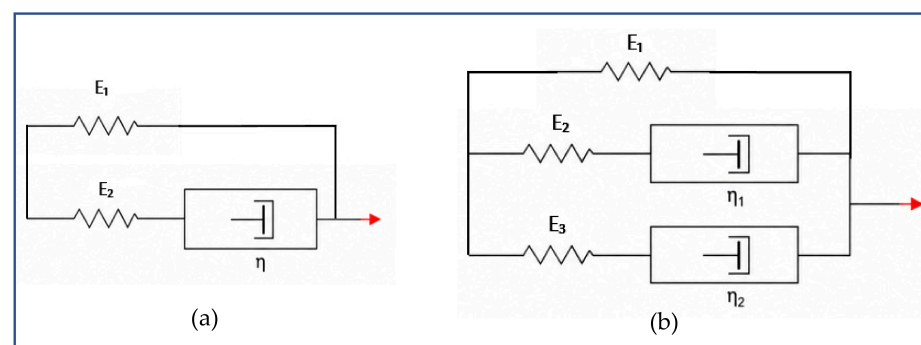


Figure 2. Viscoelastic model: (a) SLS rheological model, (b) Generalized rheological model.

The SLS model (Zener model) consists of an equilibrium spring (E_1) connected in parallel with a Maxwell element composed of a spring (E_2) and a dashpot (η). The stress response is given by:

$$\sigma(t) = \varepsilon_0 \cdot [E_1 + E_2 \cdot e^{-t/\tau}] \quad (1)$$

where:

E_1 is the equilibrium elastic modulus,

E_2 is the transient relaxation modulus,

ε_0 is the applied strain,

t is the relaxation time,

τ is the characteristic relaxation time.

The characteristic relaxation time τ was calculated as:

$$\tau = \frac{\eta}{E_2} \quad (2)$$

where:

τ is the characteristic relaxation time,

η is the viscosity coefficient,

E_2 is the elastic modulus of the Maxwell branch (MPa).

To capture the multiscale relaxation behavior of trabecular bone, a generalized Maxwell model with two viscoelastic branches was also employed:

$$\sigma(t) = \varepsilon_0 \cdot [E_1 + E_2 e^{-t/\tau_1} + E_3 e^{-t/\tau_2}] \quad (3)$$

where:

E_1 is the equilibrium elastic modulus,

E_2 and E_3 are the viscoelastic moduli,

τ_1 and τ_2 are the characteristic relaxation times associated with the fast and slow relaxation processes, respectively.

The SLS and two-branch Generalized Maxwell models were selected because they represent classical viscoelastic formulations widely used for biological tissues. While the SLS model provides a simple explanation of stress relaxation through a single characteristic relaxation phase, the Generalized Maxwell model allows for modeling several relaxation mechanisms occurring at different time scales. This study compared a single-process and a multiscale viscoelastic formulation rather than evaluating all existing rheological models.

2.5. Energy Analysis

Energy-related parameters were calculated from the fitted viscoelastic models. The initial elastic energy stored in the specimen immediately after loading was calculated as:

$$W_{init} = \frac{1}{2} (E_1 + E_2 + E_3) \varepsilon_0^2 \quad (4)$$

The long-term elastic energy retained after complete relaxation was:

$$W_{\infty} = \frac{1}{2} E_1 \varepsilon_0^2 \quad (5)$$

The dissipated energy was then determined as:

$$W_{diss} = W_{init} - W_{\infty} = \frac{1}{2} (E_2 + E_3) \varepsilon_0^2 \quad (6)$$

These parameters were used to quantify the balance between elastic energy storage and viscous dissipation in healthy and osteoarthritic trabecular bone.

2.6. Parameter Identification and Optimization

Nonlinear least-squares optimization was used to get the model parameters. To lower the chance of convergence toward local minima, several initial parameter values were examined. The highest coefficient of determination (R^2), the lowest fitting error, and the physical plausibility of the acquired parameters were used to determine the ideal solution.

The objective function was defined as:

$$SSE = \sum_{i=1}^N (\sigma_{exp,i} - \sigma_{model,i})^2 \quad (7)$$

The goodness of fit was evaluated using the coefficient of determination (R^2):

$$R^2 = 1 - \frac{\sum (\sigma_{exp} - \sigma_{model})^2}{\sum (\sigma_{exp} - \bar{\sigma}_{exp})^2} \quad (8)$$

Numerical analyses were performed using Python 3.11 with the NumPy, SciPy, and Pandas libraries.

The coefficient of determination (R^2) was used as the primary indicator of model fitting performance because all models were fitted and compared using the same experimental stress-relaxation datasets and identical testing conditions. Although error-based metrics such as RMSE can provide additional information on absolute prediction errors, R^2 was considered an appropriate comparative criterion in the present study. Future studies should incorporate complementary metrics to further assess constitutive model performance.

2.7. Statistical Analyses

SciPy (version 1.11) was used for statistical analysis. The data were expressed using the mean $\pm$ standard deviation. The Shapiro–Wilk test was used to confirm normality. The healthy and osteoarthritic groups were compared using the independent Student's t -test for normally distributed variables and the Mann–Whitney U test for non-normally distributed variables. $p < 0.05$ was established as the cutoff point for statistical significance.

2.8. Sensitivity Analysis

A local sensitivity analysis was conducted on the parameters of the Generalized Maxwell model. Each parameter (E_1 , E_2 , E_3 , τ_1 , and τ_2) was independently perturbed by $\pm 10\%$ while maintaining all other parameters constant. The resulting variations in the predicted stress-relaxation response were evaluated to identify the parameters exerting the greatest influence on model behavior and to assess the robustness of parameter estimation. This approach was selected to quantify the relative contribution of individual parameters around the optimized solution. Although global sensitivity and uncertainty analyses could provide a more comprehensive assessment of parameter interactions and model robustness, such approaches were beyond the scope of the present study and could be considered in future investigations.

2.9. Relaxation Curve Analysis

Several descriptors were extracted from each experimental relaxation curve, including the initial stress (σ_0), equilibrium stress (σ_∞), half-relaxation time ($t_{1/2}$), relaxation ratio (σ_∞/σ_0), stress–time integral, and energy-related parameters (W_{init} , W_∞ , and W_{diss}). The half-relaxation time ($t_{1/2}$) was defined as the time required for the stress to decrease by 50% of the difference between the initial stress (σ_0) and the equilibrium stress (σ_∞). These

descriptors were used to characterize the viscoelastic response and compare healthy and osteoarthritic trabecular bone specimens.

3. Results

3.1. Stress-Relaxation Behavior

Stress-relaxation tests were performed on fifteen trabecular bone specimens subjected to a constant imposed strain ($\varepsilon_0 = 0.1667$). The resulting stress–time curves exhibited a characteristic exponential decay, confirming the viscoelastic behavior of trabecular bone (Figure 3). The mean initial stress was $\sigma_0 = 2.10 \pm 0.97$ MPa, while the equilibrium stress measured at the end of the relaxation period was $\sigma_\infty = 1.11 \pm 0.74$ MPa (Table 1). The half-relaxation time was $t_{1/2} = 116.56 \pm 22.81$ s, and the relaxation ratio (σ_∞/σ_0) reached 0.48 ± 0.17 . The stress–time integral was 878.27 ± 441.13 MPa·s. Comparison between healthy and osteoarthritic (OA) specimens revealed no statistically significant differences in σ_0 , σ_∞ , relaxation ratio, or Young’s modulus ($p > 0.05$). However, the half-relaxation time tended to be lower in OA specimens (103.96 s) than in healthy specimens (125.56 s), approaching statistical significance ($p = 0.0772$).

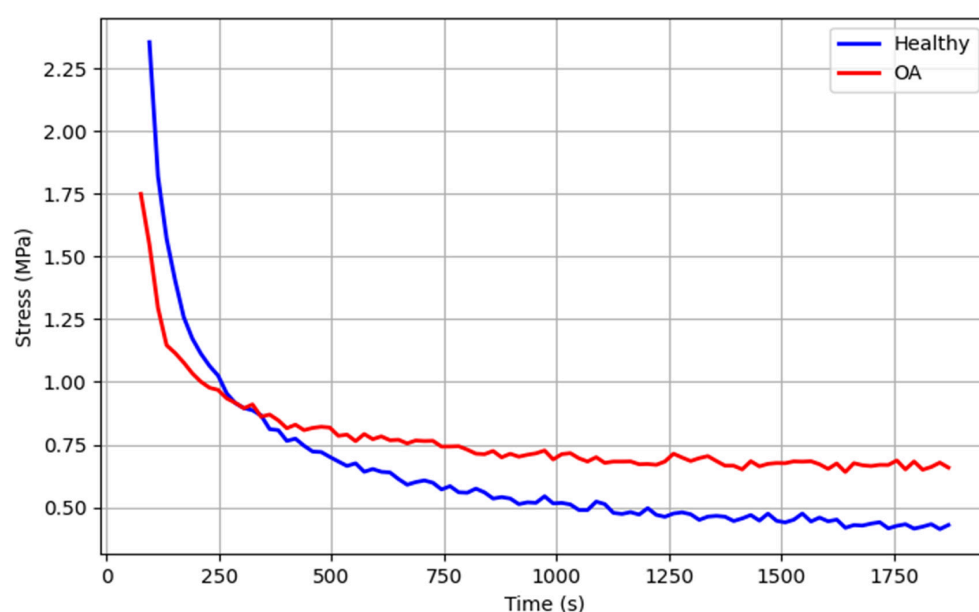


Figure 3. Evolution of stress versus time for healthy and osteoarthritic trabecular bone specimens.

Table 1. Comparison of stress-relaxation mechanical parameters between healthy and osteoarthritic human trabecular bone tissues.

Parameter (Unit)	Overall Mean $\pm$ SD	Healthy ($n = 8$) Mean	OA ($n = 7$) Mean	p -Value
Initial stress (MPa)	2.10 ± 0.97	2.12	2.07	0.931
Final stress (MPa)	1.11 ± 0.74	1.15	1.06	0.839
Half-relaxation time (s)	116.56 ± 22.81	125.56	103.96	0.077
Relaxation ratio	0.48 ± 0.17	0.468	0.499	0.738
Young’s modulus (MPa)	100.61	100.74	105.45	0.931

Mechanical parameters obtained from stress-relaxation tests in healthy and osteoarthritic (OA) human trabecular bone specimens. Values are presented as mean $\pm$ standard deviation (SD). The healthy ($n = 8$) and OA ($n = 7$) groups were compared using an independent Student’s t -test or Mann–Whitney U test according to data normality. A p -value < 0.05 was considered statistically significant.

3.2. Energy-Related Parameters

Energy parameters extracted from the relaxation curves are summarized in Table 2. The mean initial stored energy (E_{init}) was 0.175 ± 0.081 J/cm³, while the long-term retained energy (E_{long}) was 0.093 ± 0.062 J/cm³. The dissipated energy (E_{diss}) reached 0.083 ± 0.040 J/cm³. No statistically significant differences were observed between healthy and OA groups for E_{init} , E_{long} , or E_{diss} ($p > 0.05$). In contrast, the stress–time area showed a marked reduction in OA specimens (617 MPa·s) compared with healthy specimens (1065 MPa·s), with a borderline significance level ($p = 0.0599$).

Table 2. Energy dissipation characteristics derived from stress-relaxation tests in healthy and osteoarthritic human cancellous bone specimens.

Parameter	Overall Mean $\pm$ SD	Healthy ($n = 8$) Mean	OA ($n = 7$) Mean	p -Value
E_{init} (J·cm ^{−3})	0.175 ± 0.081	0.177	0.173	0.931
E_{long} (J·cm ^{−3})	0.093 ± 0.062	0.096	0.089	0.839
E_{diss} (J·cm ^{−3})	0.083 ± 0.040	0.081	0.084	0.890
Stress–time area $\sigma(t)$ (MPa·s)	878.27 ± 441.13	1064.7	617.2	0.060

Energy-related parameters calculated from stress-relaxation responses of healthy and osteoarthritic (OA) cancellous bone specimens. Values are expressed as mean $\pm$ standard deviation (SD). Statistical comparisons between groups were performed using an independent Student's t -test or Mann–Whitney U test depending on the distribution of the data. p -values indicate comparisons between healthy and OA groups. No statistically significant differences were detected between groups ($p > 0.05$).

3.3. Viscoelastic Model Fitting

The experimental relaxation curves were fitted using the SLS model and a two-branch Generalized Maxwell model. For healthy trabecular bone (Figure 4), both models reproduced the general relaxation behavior. However, the Generalized Maxwell model provided a substantially better fit ($R^2 = 0.998$) than the SLS model ($R^2 = 0.949$), particularly during the intermediate and long-term relaxation phases. Similar observations were obtained for OA specimens (Figure 5). The Generalized Maxwell model accurately captured the complete relaxation response ($R^2 = 0.993$), whereas the SLS model showed larger deviations from the experimental data ($R^2 = 0.939$).

3.4. Viscoelastic Parameters

The parameters identified using the SLS model are presented in Table 3. The equilibrium modulus (E_1) remained similar between groups, whereas the transient modulus (E_2) tended to be higher in OA specimens. The relaxation time τ showed a tendency to decrease in OA bone (59.8 s) compared with healthy bone (124.1 s), although the difference was not statistically significant. The parameters obtained from the Generalized Maxwell model are reported in Table 4. OA specimens exhibited higher values of E_2 and E_3 , shorter τ_1 , and longer τ_2 compared with healthy specimens. However, none of these differences reached statistical significance ($p > 0.05$).

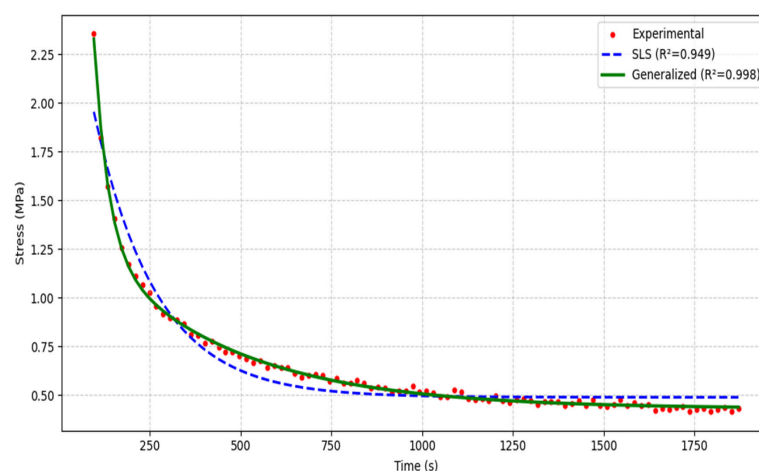


Figure 4. Representative experimental stress–relaxation curve of a healthy tibial trabecular bone specimen (red points) compared with the SLS model (blue dashed line) and the two-branch Generalized Maxwell model (green curve). The SLS model provides a moderate fit ($R^2 = 0.949$) but underestimates the mid-relaxation region, whereas the Generalized Maxwell model more accurately captures the entire relaxation behavior ($R^2 = 0.998$).

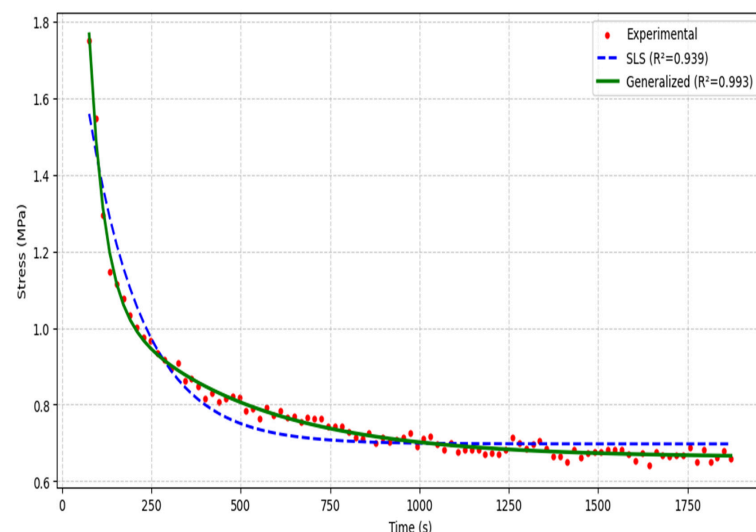


Figure 5. Representative experimental stress–relaxation curve of an osteoarthritic tibial trabecular bone specimen (red points) compared with the SLS model (blue dashed line) and the two-branch Generalized Maxwell model (green curve). The SLS model provides a moderate fit ($R^2 = 0.939$) but underestimates the mid-relaxation region, whereas the Generalized Maxwell model more accurately captures the entire relaxation behavior ($R^2 = 0.993$).

Table 3. SLS Model parameters for healthy vs. osteoarthritic trabecular bone.

Parameter	Healthy ($n = 8$) Mean	OA ($n = 7$) Mean	p -Value
E_1 (MPa)	1.187	1.101	0.8410
E_2 (MPa)	2.625	4.246	0.2538
τ (s)	124.050	59.822	0.1218
R^2	0.923	0.922	

Viscoelastic parameters obtained from the Standard Linear Solid (SLS) model fitted to stress-relaxation responses of healthy and osteoarthritic (OA) human trabecular bone specimens. Values are expressed as mean values for each group. Statistical comparisons between healthy ($n = 8$) and OA ($n = 7$) groups were performed using an independent Student's t -test or Mann–Whitney U test depending on data distribution. A p -value < 0.05 was considered statistically significant. R^2 values represent the goodness of fit of each model and were not statistically compared between groups.

Table 4. Generalized Maxwell Model (Two-Branch) Parameters.

Parameter (Unit)	Healthy ($n = 8$) Mean	OA ($n = 7$) Mean	p -Value
E_1 (MPa)	1.128	0.874	0.5909
E_2 (MPa)	81.135	264.314	0.1922
τ_1 (s)	26.400	17.365	0.1179
E_3 (MPa)	0.560	0.840	0.1605
τ_2 (s)	324.856	1432.495	0.4294
R^2	0.981	0.970	

Viscoelastic parameters derived from the two-branch Generalized Maxwell model fitted to stress-relaxation curves of healthy and osteoarthritic (OA) human trabecular bone specimens. Values are expressed as mean values for healthy ($n = 8$) and OA ($n = 7$) groups. Statistical comparisons were performed using an independent Student's t -test or Mann–Whitney U test according to data distribution. Statistical significance was set at $p < 0.05$. R^2 values represent the goodness of fit of each model and were not statistically compared between groups.

3.5. Sensitivity Analysis

Sensitivity analyses of the Generalized Maxwell model are shown in Figures 6 and 7. For healthy trabecular bone, variations of $\pm 10\%$ in the viscoelastic parameters modified the predicted relaxation response while preserving the overall curve shape. Changes in E_1 primarily influenced the early relaxation phase, whereas τ_2 had the strongest effect on the long-term response. For OA trabecular bone, parameter perturbations generated larger deviations from the baseline response, indicating increased model sensitivity. Variations in E_1 and τ_1 mainly affected the initial relaxation stage, while E_2 , E_3 , and τ_2 controlled the intermediate and long-term relaxation behaviour.

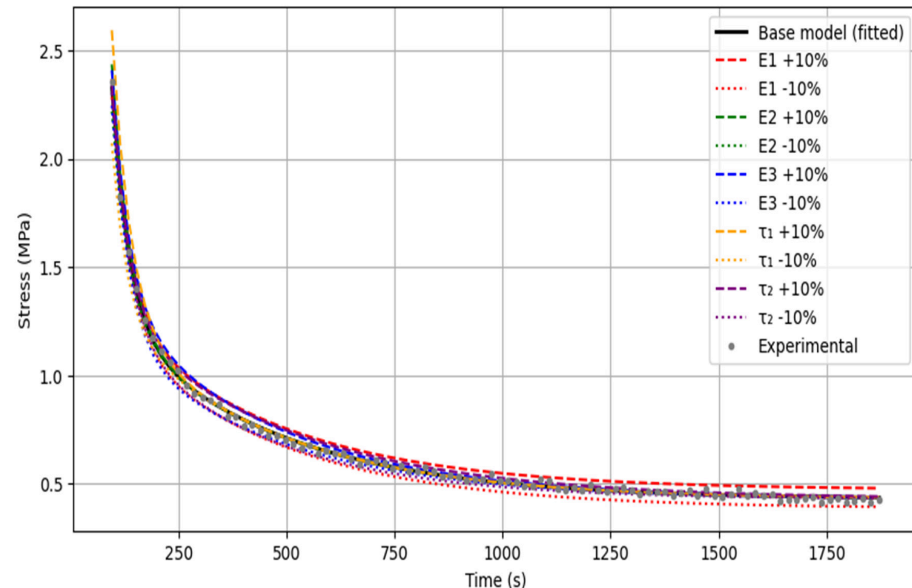


Figure 6. Sensitivity analysis of the generalized Maxwell model applied to healthy trabecular bone. The solid black line represents the fitted baseline model, and the grey dots show the experimental stress-relaxation data. Colored dashed curves illustrate the effect of $\pm 10\%$ variations in viscoelastic parameters (E_1 , E_2 , E_3 , τ_1 , τ_2) on the predicted stress response.

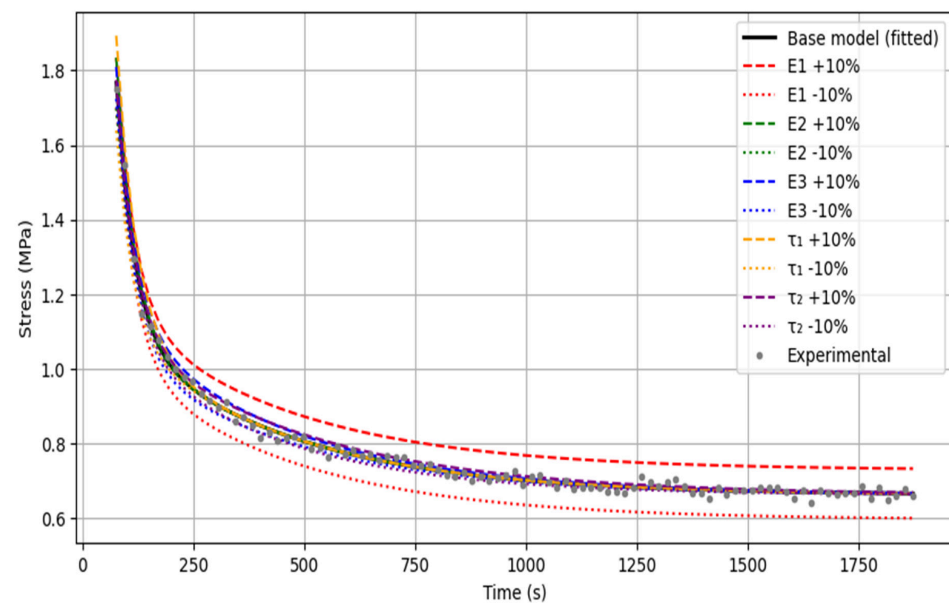


Figure 7. Sensitivity analysis of the generalized Maxwell model applied to osteoarthritic trabecular bone. The solid black line represents the fitted baseline model, and the grey dots show the experimental stress-relaxation data. Colored dashed curves illustrate the effect of $\pm 10\%$ variations in viscoelastic parameters (E_1 , E_2 , E_3 , τ_1 , τ_2) on the predicted stress response. Changes in E_1 and τ_1 dominate the early relaxation phase (0–200 s), whereas variations in E_2 , E_3 , and τ_2 primarily affect the mid- to long-term response (200–1800 s). The broader spread of perturbed curves reflects the increased mechanical sensitivity and heterogeneity characteristic of osteoarthritic trabecular bone.

4. Discussion

Stress-relaxation tests confirmed the time-dependent viscoelastic behavior of human trabecular bone. No statistically significant differences were detected in the investigated mechanical parameters between healthy and osteoarthritic groups (all $p > 0.05$). Although OA specimens exhibited numerically shorter relaxation times, these observations should be considered exploratory and require confirmation in larger cohorts. The absence of statistically significant differences in elastic parameters is consistent with previous studies reporting that OA-related bone changes are often associated with microarchitectural remodeling rather than major alterations in intrinsic tissue properties [13,18,19]. Previous studies have suggested that subchondral sclerosis, trabecular thickening, altered connectivity, and changes in pore architecture may influence fluid redistribution and load transfer mechanisms without substantially modifying the apparent elastic response [18]. However, these microstructural and fluid-related mechanisms were not directly evaluated in the present study.

Previous biomechanical investigations have similarly highlighted the strong dependence of trabecular mechanical behavior on architectural organization, anisotropy, and local structural heterogeneity rather than on tissue-level elastic properties alone [8,20].

The energy analysis demonstrated that trabecular bone combines elastic energy storage with viscous energy dissipation. Although the stress–time integral was numerically lower in OA specimens, this difference was not statistically significant. Therefore, no conclusions regarding inter-group differences can be drawn from the present data, and these observations require validation in larger studies. Previous studies have shown that the ability of trabecular bone to dissipate mechanical energy plays a crucial role in shock absorption and protection of the osteochondral unit under repetitive loading conditions [19–21]. Therefore, variations in energy dissipation may influence mechanical transfer to the overlying cartilage, although their contribution to joint degeneration remains to be confirmed [5,22].

Furthermore, recent imaging and computational studies have demonstrated that local variations in trabecular architecture strongly influence stress redistribution, energy absorption, and damage accumulation within osteoarthritic bone [23,24].

The Generalized Maxwell model consistently outperformed the SLS model in both healthy and OA groups. This result indicates that trabecular bone relaxation cannot be fully described by a single relaxation process and is better represented by multiple characteristic time scales [25–27]. Based on previous studies, the identified short- and long-term relaxation mechanisms have been associated with contributions from fluid flow, collagen–mineral interactions, and microstructural rearrangements [28,29]. However, these mechanisms were not directly quantified in the present work. Similar multiscale relaxation behavior has been reported in bone and cartilage tissues, where previous studies have suggested contributions from both matrix viscoelasticity and fluid transport to stress decay [28,29]. Direct assessment of these mechanisms was beyond the scope of the present study.

The Generalized Maxwell model parameters showed substantial variability, particularly the equilibrium-related modulus E_2 . Although the mean E_2 value was numerically higher in the OA group, no statistically significant difference was detected. The observed variability likely reflects the heterogeneous nature of human trabecular bone, including differences in microarchitecture, tissue composition, and remodeling patterns. Therefore, these parameters should be interpreted as descriptive indicators of viscoelastic behavior rather than definitive markers of osteoarthritic alteration.

Although no statistically significant differences in viscoelastic behavior were identified between groups, the multiscale parameters obtained from the Generalized Maxwell model provide useful descriptors of subchondral bone mechanical behavior. Their potential role in load transmission to the overlying cartilage and osteoarthritis progression remains to be established in larger studies. Together with previous evidence, these findings support the continued investigation of subchondral bone as a contributor to osteoarthritis progression and a potential therapeutic target alongside cartilage-focused interventions [22,26,27]. Although additional validation is required, combining mechanical testing with constitutive modeling may help improve the biomechanical characterization of osteoarthritic tissues and support future diagnostic and therapeutic developments targeting the osteochondral unit [24–27].

The identified viscoelastic parameters should currently be considered biomechanical descriptors rather than clinical biomarkers. Their potential clinical relevance requires further validation through larger studies integrating imaging, clinical data, and longitudinal assessment of osteoarthritis progression.

Recent studies have further highlighted that osteoarthritis-related alterations in subchondral bone involve complex changes in trabecular architecture, mineralization, and tissue-level mechanical behavior. These microstructural modifications have been reported in previous studies to influence stress distribution, fluid transport, and time-dependent mechanical responses of trabecular bone. However, the relationship between structural remodeling and viscoelastic properties remains incompletely understood, particularly in human osteoarthritic specimens.

Several limitations should be acknowledged. First, the sample size was relatively small, which may have limited the statistical power of group comparisons and the detection of subtle differences between healthy and osteoarthritic specimens. Due to the limited availability of human tibial trabecular bone samples, particularly osteoarthritic specimens, increasing the cohort size was not feasible in the current study. Therefore, the observed numerical differences should be interpreted cautiously and require confirmation in future investigations involving larger cohorts. Donor-related factors, including age, sex, and bone

mineral density, may influence trabecular bone mechanical properties and represent potential confounding variables. However, due to the limited sample size, multivariate statistical approaches accounting for these factors were not feasible. Future studies with larger cohorts should consider these variables to better isolate disease-related biomechanical changes. Second, microstructural characterization using techniques such as micro-computed tomography or histological analysis was not performed. Consequently, direct relationships between trabecular architecture, tissue remodeling, and viscoelastic parameters could not be established. Similarly, fluid transport properties and poromechanical behavior were not measured; therefore, any interpretation involving fluid-related mechanisms should be considered hypothesis-generating and based on previous literature rather than direct experimental evidence from this study. Future investigations combining mechanical testing with detailed structural characterization are required to better understand the mechanisms underlying osteoarthritis-related biomechanical changes. Third, all tests were conducted *ex vivo* under controlled laboratory conditions. Although specimens were tested on the day of harvesting to minimize tissue degradation and preserve their mechanical properties, variations in hydration state and biological preservation cannot be completely excluded and may influence the measured viscoelastic response. The theoretical formulations of the SLS and Generalized Maxwell models assume an ideal step-strain input. In the present study, the target strain was achieved through a finite loading ramp. Consequently, the identified parameters should be interpreted as effective relaxation descriptors rather than absolute intrinsic material constants. Nevertheless, because all specimens were tested under identical loading conditions, the comparative analysis between healthy and osteoarthritic groups remains consistent. In addition, effect size estimates and confidence intervals were not included in the present analysis. Future studies with larger cohorts should incorporate these complementary statistical approaches to better quantify the magnitude and uncertainty of biomechanical differences between groups.

Finally, the SLS and Generalized Maxwell models are theoretically derived for an ideal step-strain input [28,29]. In the present study, the imposed strain was reached through a finite loading ramp (0.5 mm/min corresponding to approximately 180 s). Consequently, the identified viscoelastic parameters should be interpreted as effective relaxation descriptors rather than intrinsic material constants. Nevertheless, because all specimens were tested under identical loading conditions, the comparative analysis between healthy and osteoarthritic groups remains valid. Similar limitations have been discussed in previous studies involving biological tissues subjected to finite-rate loading protocols [30,31].

5. Conclusions

In this study, the viscoelastic behavior of healthy and osteoarthritic human trabecular bone was investigated using stress-relaxation experiments and constitutive modeling based on the Standard Linear Solid (SLS) and two-branch Generalized Maxwell models. The experimental findings enabled the characterization of viscoelastic properties, including elastic moduli, relaxation times, and energy-related descriptors, and confirmed the time-dependent mechanical behavior of human trabecular bone.

No statistically significant differences were detected between healthy and osteoarthritic specimens for the investigated mechanical and viscoelastic parameters. Although numerical differences were observed in stress-relaxation behavior and energy-related descriptors, these findings should be considered exploratory and cannot be interpreted as evidence of disease-related biomechanical differences. Further investigations involving larger cohorts and complementary structural and fluid-related characterization are required to better elucidate the relationships between trabecular microarchitecture, fluid–solid interactions, and viscoelastic behavior.

The Generalized Maxwell model provided the most accurate description of the experimental data among the models examined, highlighting the multiscale character of trabecular bone relaxation behavior. Additionally, the sensitivity analysis identified equilibrium stiffness and relaxation times as influential parameters affecting the predicted mechanical response, emphasizing their relevance for the characterization of bone viscoelasticity.

This work provides an integrated evaluation of human trabecular bone under healthy and osteoarthritic conditions using experimental, energy-based, numerical, and sensitivity analyses. The proposed experimental and modeling framework contributes to the biomechanical characterization of subchondral bone and provides quantitative descriptors that may support future biomechanical investigations. However, given the exploratory nature of this study and its limited sample size, further research involving larger and more balanced cohorts, complementary microstructural characterization using micro-computed tomography, and advanced poroviscoelastic formulations is required to better understand the relationships between trabecular architecture, fluid transport mechanisms, and viscoelastic behavior during osteoarthritis progression.

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Institutional Review Board Statement: This study was approved by the Ethics Committee of La Rabta University Hospital (Tunis, Tunisia) on 19 October 2018 and was conducted in accordance with the principles of the Declaration of Helsinki. Written informed consent was obtained from all participants prior to tissue collection. Human trabecular bone specimens were collected from adult patients undergoing knee arthroplasty surgery.

Informed Consent Statement: Written informed consent was obtained from all participants prior to tissue collection. As the study involved the analysis of anonymized human trabecular bone specimens and no identifiable patient information is reported, specific consent for publication was not applicable.

Data Availability Statement: The data supporting the findings of this study are available from the corresponding authors upon reasonable request.

Conflicts of Interest: The authors declare no conflicts of interest.

References

1. GBD 2021 Osteoarthritis Collaborators. Global, Regional, and National Burden of Osteoarthritis, 1990–2020 and Projections to 2050. *Lancet Rheumatol.* **2023**, *5*, e508–e522.
2. Castro-Dominguez, F.; Tibesku, C.O.; McAlindon, T.E.; Freitas, R.; Ivanavicius, S.; Kandaswamy, P.; Sears, A.; Latourte, A. Literature Review to Understand the Burden and Current Non-Surgical Management of Moderate–Severe Pain Associated with Knee Osteoarthritis. *Rheumatol. Ther.* **2024**, *11*, 1457–1499. [[CrossRef](#)] [[PubMed](#)]
3. Hunter, D.J.; March, L.; Chew, M. Osteoarthritis in 2020 and Beyond: A Lancet Commission. *Lancet* **2020**, *396*, 1711–1712. [[CrossRef](#)] [[PubMed](#)]
4. Savvari, P.; Skiadas, I.; Barmpouni, M.; Papadakis, S.A.; Psychogios, V.; Pastroudis, A.P.; Skarpas, G.A.; Tsoutsanis, A.; Garofalakis, A.; Katsifis, G.; et al. Moderate to Severe Osteoarthritis: What Is the Economic Burden for Patients and the Health Care System? Insights from the “PONOS” Study. *Cartilage* **2024**, *15*, 268–277. [[CrossRef](#)] [[PubMed](#)]
5. Hu, Y.; Chen, X.; Wang, S.; Jing, Y.; Su, J. Subchondral Bone Microenvironment in Osteoarthritis and Pain. *Bone Res.* **2021**, *9*, 20. [[CrossRef](#)] [[PubMed](#)]
6. Fan, X.; Wu, X.; Crawford, R.; Xiao, Y.; Prasad, I. Osteochondral Interface Changes in Osteoarthritis: A Review. *Front. Cell Dev. Biol.* **2021**, *9*, 659654. [[CrossRef](#)] [[PubMed](#)]
7. Oftadeh, R.; Perez-Viloria, M.; Villa-Camacho, J.C.; Vaziri, A.; Nazarian, A. Biomechanics and Mechanobiology of Trabecular Bone: A Review. *J. Biomech.* **2015**, *48*, 832–841.

8. Li, G.; Yin, J.; Gao, J.; Cheng, T.S.; Pavlos, N.J.; Zhang, C.; Zheng, M.H. Subchondral Bone in Osteoarthritis: Insight into Risk Factors and Microstructural Changes. *Arthritis Res. Ther.* **2013**, *15*, 223. [[CrossRef](#)] [[PubMed](#)]
9. Oei, E.H.G.; Hirvasniemi, J.; van Zadelhoff, T.A.; van der Heijden, R.A. Osteoarthritis Year in Review 2021: Imaging. *Osteoarthr. Cartil.* **2022**, *30*, 226–236. [[CrossRef](#)] [[PubMed](#)]
10. Pahr, D.H.; Zysset, P.K. Recent Advances in Constitutive Modelling of Bone Tissue. *J. Mech. Behav. Biomed. Mater.* **2020**, *110*, 103858.
11. Micheletti, C.; Shah, F.A. Bone Hierarchical Organization through the Lens of Materials Science: Present Opportunities and Future Challenges. *Bone Rep.* **2024**, *22*, 101783. [[CrossRef](#)] [[PubMed](#)]
12. Mow, V.C.; Kuei, S.C.; Lai, W.M.; Armstrong, C.G. Biphasic Creep and Stress Relaxation of Articular Cartilage in Compression. *J. Biomech. Eng.* **1980**, *102*, 73–84. [[CrossRef](#)] [[PubMed](#)]
13. Ebrahimi, M.; Finnilä, M.A.J.; Turkiewicz, A.; Englund, M.; Saarakkala, S.; Korhonen, R.K.; Tanska, P. Elastic, Dynamic Viscoelastic and Model-Derived Fibril-Reinforced Poroelastic Mechanical Properties of Normal and Osteoarthritic Human Femoral Condyle Cartilage. *Ann. Biomed. Eng.* **2021**, *49*, 2622–2634. [[CrossRef](#)] [[PubMed](#)]
14. Findley, W.N.; Lai, J.S.; Onaran, K. *Creep and Relaxation of Nonlinear Viscoelastic Materials*; Dover Publications: New York, NY, USA, 1989.
15. Wu, B.; Yuan, L.; Liu, M.; Tang, M.; Jiang, D.; Yi, Y.; Ma, S.; Yan, B.; Lu, Y. Construction of a Viscoelastic Model of Human Cancellous Bone in Alveolar Bone Based on Bone Mineral Density Distribution. *Materials* **2023**, *16*, 7427. [[CrossRef](#)] [[PubMed](#)]
16. Liu, Z.; Ling, S.D.; Liang, K.; Chen, Y.; Niu, Y.; Sun, L.; Li, J.; Du, Y. Viscoelasticity of ECM and Cells—Origin, Measurement and Correlation. *Mechanobiol. Med.* **2024**, *2*, 100082. [[CrossRef](#)] [[PubMed](#)]
17. Khokhar, K.; Conaghan, P.G. Bone in Osteoarthritis: Imaging and Interventions. *Curr. Opin. Rheumatol.* **2022**, *34*, 73–78. [[CrossRef](#)] [[PubMed](#)]
18. Xiong, Z.; Bensidhoum, M.; Ferry, D.; Huang, X.; Potier, E.; Hoc, T. Multiscale Modeling of Viscoelastic Behavior in Demineralized Human Trabecular Bone. *Int. J. Mech. Sci.* **2025**, *307*, 110887. [[CrossRef](#)]
19. Goldring, S.R.; Goldring, M.B. Changes in the Osteochondral Unit During Osteoarthritis: Structure, Function and Cartilage–Bone Crosstalk. *Nat. Rev. Rheumatol.* **2016**, *12*, 632–644. [[CrossRef](#)] [[PubMed](#)]
20. Hernandez, C.J.; Majumdar, S. Osteoarthritis and Bone: A Complex Interaction. *Nat. Rev. Rheumatol.* **2023**, *19*, 45–59.
21. Maquer, G.; Musy, S.N.; Wandel, J.; Zysset, P.K. Bone Microstructure Determines the Energy Absorption Capacity of Trabecular Bone. *J. Mech. Behav. Biomed. Mater.* **2020**, *101*, 103440.
22. Burr, D.B.; Gallant, M.A. Bone Remodelling in Osteoarthritis. *Nat. Rev. Rheumatol.* **2012**, *8*, 665–673. [[CrossRef](#)] [[PubMed](#)]
23. Findlay, D.M.; Atkins, G.J. Osteoblast–Chondrocyte Interactions in Osteoarthritis. *Curr. Osteoporos. Rep.* **2014**, *12*, 127–134. [[CrossRef](#)] [[PubMed](#)]
24. Ma, C.; Du, T.; Niu, X.; Fan, Y. Biomechanics and Mechanobiology of the Bone Matrix. *Bone Res.* **2022**, *10*, 59. [[CrossRef](#)] [[PubMed](#)]
25. Muratovic, D.; Findlay, D.M.; Quinn, M.J.; Quarrington, R.D.; Solomon, L.B.; Atkins, G.J. Microstructural and Cellular Characterisation of the Subchondral Trabecular Bone in Human Knee and Hip Osteoarthritis Using Synchrotron Tomography. *Osteoarthr. Cartil.* **2023**, *31*, 1224–1233. [[CrossRef](#)] [[PubMed](#)]
26. Pahr, D.H.; Reisinger, A.G. A Review on Recent Advances in the Constitutive Modeling of Bone Tissue. *Curr. Osteoporos. Rep.* **2020**, *18*, 696–704. [[CrossRef](#)] [[PubMed](#)]
27. Hamed, E.; Heinemann, F.; Schwiedrzik, J.; Zysset, P.K. A Two-Layer Elasto-Visco-Plastic Rheological Model for the Material Parameter Identification of Bone Tissue. *Biomech. Model. Mechanobiol.* **2020**, *19*, 2149–2162. [[CrossRef](#)] [[PubMed](#)]
28. García-Aznar, J.M.; Nasello, G.; Hervas-Raluy, S.; Pérez, M.A.; Gómez-Benito, M.J. Multiscale Modeling of Bone Tissue Mechanobiology. *Bone* **2021**, *152*, 116032. [[CrossRef](#)] [[PubMed](#)]
29. Crolla, J.P.; Lawless, B.M.; Cederlund, A.A.; Aspden, R.M.; Espino, D.M. Analysis of Hydration and Subchondral Bone Density on the Viscoelastic Properties of Articular Cartilage. *BMC Musculoskelet. Disord.* **2022**, *23*, 228. [[CrossRef](#)] [[PubMed](#)]
30. Hunter, D.J.; Bierma-Zeinstra, S.M.A. Osteoarthritis. *Lancet* **2019**, *393*, 1745–1759. [[PubMed](#)]
31. Oláh, T.; Cucchiari, M.; Madry, H. Temporal Progression of Subchondral Bone Alterations in OA Models Involving Induction of Compromised Meniscus Integrity in Mice and Rats: A Scoping Review. *Osteoarthr. Cartil.* **2024**, *32*, 1220–1234. [[CrossRef](#)] [[PubMed](#)]

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