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Original Article

Dynamic Bone Quality: A Noninvasive Measure of Bone's Biomechanical Property in Osteoporosis

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Abstract

We describe a novel approach to characterize bone quality noninvasively, a measurement that quantifies aggregate shock-absorption capacity of load-bearing bones as a measure of mechanical structural integrity during exposure to real-time self-induced in vivo loading associated with heel strike. The outcome measure, damping factor, was estimated at 5 load-bearing anatomical sites: ankle, tibial tuberosity, femoral condyle, lower back (at 3rd lumbar vertebra), and upper back (7th thoracic vertebra) plus the forehead in 67 patients with postmenopausal osteoporosis with and without documented vertebral fractures. The damping value was significantly lower in patients with vertebral fractures compared with those without a fracture (range: −36% to −72%; median: −44%). In these women with osteoporosis, damping factor was able to discriminate between patients with and without vertebral fractures, whereas traditional measures of bone density and biomechanical measures obtained from bone geometry were not significantly different between the groups.

Key Words: Damping; dynamic bone quality; fracture; osteoporosis; shock absorption.

Introduction

The objective of this study was to further develop and evaluate an inexpensive noninvasive screening tool as a supplement to bone mineral density (BMD) measurements for discriminating women with postmenopausal osteoporosis who had vertebral fracture from those without any fracture.

The incidence of osteoporosis is higher in women than in men. About 50% of women older than 50 yr will develop osteoporosis during their remaining lives *(1)*. The ''gold standard'' for the prefracture diagnosis of osteoporosis is measurement of BMD by dual-energy X-ray absorptiometry (DXA). BMD correlates better with fracture risk than with cholesterol with heart disease and blood pressure with stroke. However, although patients with osteoporosis are at high risk of fracture, most patients with ''osteoporosis'' by DXA criteria do not fracture, and most fractures occur in patients who do not meet the DXA criteria for osteoporosis. There are other determinants of bone strength and fracture risk besides BMD, such as microarchitecture, mechanical structural integrity, and bone turnover *(2–6)*. Thus, although BMD measurement by DXA is a useful test, having complimentary measures of bone quality or bone strength should provide additional information to assess fracture risk *(7–11)*.

We describe a complimentary measure of bone quality by evaluating bone shock absorption (BSA), damping factor (ζ), as an aggregate response of bone's structural integrity under self-induced dynamic ''realistic'' in vivo loading. Structural failure of human bone and/or any mechanical system rarely occurs under unloaded or static conditions. Therefore, better understanding of bone fracture and prevention requires measurement of both static and dynamic biomechanical properties of bone when exposed to realistic ''real world'' in vivo external loadings *(12,13)*. The importance of understanding dynamic mechanical behavior of bone has been discussed in

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the literature, but so far, no study has presented a noninvasive tool for quantifying this property in humans. The BSA measurement is designed to capture "dynamic bone quality" properties noninvasively, as measured by damping factor (ζ), whereas traditional techniques, such as DXA, provide information about static bone quality properties. Osteoporosis is associated with decreased bone mass (as reflected by decrease in BMD and bone mineral content) and deterioration of trabecular architecture, which collectively cause detrimental impact on the quality of bone, that is, increased stiffness and brittleness and decreased strength *(2,3)*. Among all natural shock absorbers in the human body, trabecular bone has the greatest capability (170 times higher than that provided by cartilage and synovial fluid) to attenuate the incoming shock wave associated with heel strike during walking and running *(14–17)*. In healthy subjects, 70% of the incoming shock waves associated with natural daily activity of heel strikes are absorbed by the body's natural shock absorbers before reaching the forehead *(18–20)*. Previous researchers have suggested that cumulative everyday cyclic loading activities may give rise to microdamage in the form of microcracks, thereby affecting the mechanical property of bone, toughness, that is, its shock-absorbing capacity *(14,21–23)*. Because trabecular bones are detrimentally affected by osteoporosis, the natural shock-absorbing capacity will be compromised.

The current study describes the measurement of the BSA capacity of individuals with postmenopausal osteoporosis during exposure to realistic in vivo loading associated with heel strike. BSA evaluation provides a "signature" of dynamic bone quality, because the measurement of shock-absorption capacity captures an aggregate response of the load-bearing bone because of dynamic loading associated with heel strike.

The main focus of this study was to determine if the information provided by BSA evaluation could serve as a supplementary tool (along with BMD) to better discriminate patients with postmenopausal osteoporosis with and without fractures. Noninvasive measures of natural shock absorbers of musculoskeletal systems may be potentially effective and economical descriptors of clinical and preclinical statuses of degenerative musculoskeletal diseases in elderly individuals *(15,24–26)*. The BSA evaluation approach is based on sound theories in engineering, physics, mechanics, and physiology. The BSA technique is totally noninvasive, objective, simple, and quick to administer *(25–27)*. Preliminary studies by our group and those by others have shown that natural shock absorbers of the musculoskeletal system are impaired in osteoarthritic patients, but there is no information regarding how osteoporosis affects natural shock absorption, thus forming the rationale for conducting the current study *(22,23,25–27)*.

Materials and Methods

Subjects

Sixty-seven Caucasian women aged between 65 and 86 yr were recruited from the clinical practice of a coinvestigator (N.B.W.) at the Bone Health and Osteoporosis Center at the University of Cincinnati. All subjects met the World Health Organization BMD criterion for osteoporosis—T-score: 2.5 standard deviation (SD) or more below the mean values of young healthy adult females in the spine or hip *(1)*. Eligibility criteria required subjects to have at least 2 vertebrae from L1 through L4 unaffected by degenerative change or fracture. Subjects were physically and mentally able to perform the heel strike maneuver, able to walk at a moderate pace for 5 min without cane or other assistance, and lift and hit their foot 5 times while standing on the other foot unassisted. Subjects who had factors that might confound the results of the heel strike test, including significant arthritis in the hip or knee that limited ambulation; fracture of a lower extremity bone within the last 2 yr; or surgery to the spine, hip, knee, or ankle, were excluded.

One investigator (N.B.W.) was aware of the subjects' fracture status but was blinded to the BSA test results, and the remaining investigators were aware of the BSA test results but blinded about the subjects' fracture status. The protocol was approved by the University Institutional Review Board, and all subjects gave informed consent.

Bone Densitometry and Fracture Assessment

BMD of 34 patients was determined using DXA a Hologic Delphi device (version 11.2:30, Hologic Inc., Waltham, MA), and the remaining 33 subjects were tested with a GE-Lunar Prodigy (version 8.8. DF + 13153, GE Medical Systems, Madison, WI). Coefficients of variations in adults for both devices were 0.8% in the spine and approximately 1–2% in the hip regions. BMD values obtained from GE Lunar device were converted to Hologic-equivalent values using equations developed at our center based on cross-calibration of GE and Hologic equipment. Vertebral fractures were assessed from lateral spine images acquired using the DXA device or lateral spine X-rays; reduction in anterior, middle, and/or posterior vertebral height by 20% or more constituted a fracture *(28)*. Twenty-eight of the total subjects had 1 or more vertebral fractures, whereas 39 subjects had no vertebral fractures.

Static Bone Quality Properties

Biomechanical properties of bone under unloaded or static condition were calculated from bone density and bone geometry data obtained from DXA for the femoral neck. Because the empirical equations used for the calculation of biomechanical measures were based on data collected with Hologic device, only a subset of our data (n = 34) was used for this calculation *(29)*. Biomechanical measures were calculated to capture bone's bending resistance, cortical instability, and compressive strength. These biomechanical variables include section modulus, buckling ratio, cortical thickness, and compressive strength. Formulae to calculate these variables were obtained from Riancho et al *(29)*.

Dynamic Bone Response Properties—Measurement Tool and Experimental Task

Low-mass, skin-mounted accelerometers were attached to the bony prominences by metal/hard Lexican holders secured on the skin with micropore tape at the following 5 load-bearing anatomical sites: ankle, tibial tuberosity, femoral condyle, lower back (at the 3rd lumbar vertebra), upper back (7th thoracic vertebra), plus forehead (Fig. 1). Previous studies by our group *(25,26)* and others *(30,31)* have shown that the use of low-mass accelerometers (i.e., increasing its resonance frequency) minimizes the effect of soft tissue under the accelerometers ($<5\%$ loss of accelerometer information), thereby providing a reliable measure of shock wave propagation during heel strike. A previous study *(32)* showed that the whole-bone strength is influenced by both the trabecular structure and the cortical bone surrounding the trabecular structure. Therefore, in this study, accelerometers were attached to the bony prominences of the whole bone, giving an aggregate damping value for both trabecular and cortical bones. The signals from accelerometers were sampled at 640 Hz with a 16-bit analog to digital convertor board equipped with sample and hold circuit.

The subjects completed stationary foot-striking tasks in our gait laboratory fitted with a force platform system (Model OR6 AMTI force platform system, Advance Mechanical Technology Inc., Watertown, MA). The stationary task consisted of lifting the foot (while keeping the other foot stationary on the ground) and placing down with the heel striking the force platform with a force equivalent to that used during natural walking. This stationary task was repeated 5 times. The results from 5 foot strikes were averaged for statistical analysis. During testing, all subjects wore full-body safety harness.

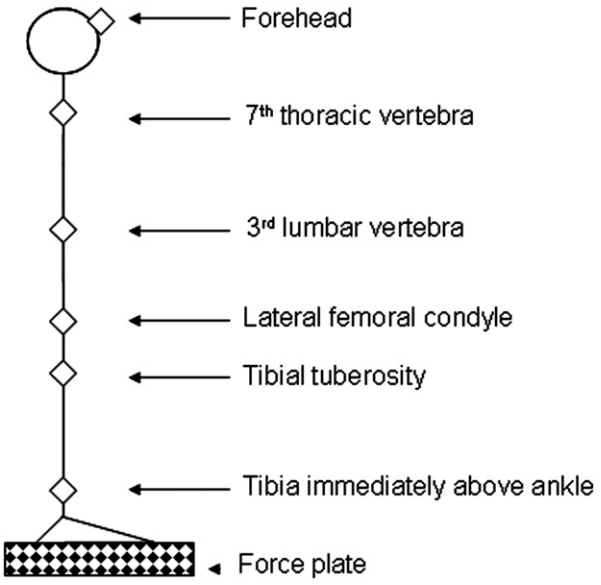

Fig. 1. Schematic of accelerometer-placement sites.

Dynamic Bone Quality Measure: Measurement of Bone Shock Absorption

Bone damping (ζ) and resonance/dominant frequencies at the anatomical sites were calculated assuming the musculoskeletal system being modeled as a single degree of freedom system responding to the transient force caused by the heel strike in accordance with the solution to the second-order differential equation *(33–36)*. The second-order differential equation solution may take the form of a frequency response function (FRF) or transfer function between force plate and the acceleration at the measurement site. The advantage of this form is that the equation is known explicitly, and the FRF is obtained experimentally. Thus, ζ was obtained directly from the measured FRF using the structural bandwidth and resonance frequency method *(34,35)*. The structural bandwidth is manifest in the FRF real part as the frequency separation between 2 extremes, symmetric about the resonance frequency, whereas the resonance frequency coincides with a single peak in the FRF imaginary part. The ζ is calculated as the ratio: 1/2 the structural bandwidth divided by the resonance frequency. Using the aforementioned structural bandwidth and resonance frequency method, ζ was obtained directly from a measured FRF *(34,35)*. We have used these variables successfully in our earlier studies in adults and adolescents *(25,37)*. As waveforms vary because of the frequency content of the transient force and transient acceleration, the ratios of heel strike force and the accelerations can only be consistent on a frequency-by-frequency basis, and this is the reason for using the frequency domain or FRF analysis. Another advantage of conducting the analysis in frequency domain is that it permits the calculation of the damping value independent of surface and force-time peak level. Test and retest of BSA trial was carried out in our earlier study, where the mean coefficients of variations were 5% and 8.6% for resonant frequency and damping, respectively *(37)*.

Data Analysis

All study-related data were checked for transcriptional errors by comparing against the original data. The statistical software used for the purpose of advanced analyses was SPSS (Statistical Package for Social Sciences, SPSS Inc., IBM Company, Chicago, IL) version 11.5. Descriptive data analysis summarized the data in terms of certain descriptive statistics, such as mean, median, SD, skewness, kurtosis, range, and others. Box plots were generated for each continuous variable to identify potential outliers and extreme observations for each variable. This was done separately for the 2 groups (fracture and nonfracture patients). The influence of outliers and extreme observations were assessed by conducting all statistical analyses ''with'' and ''without'' such observations in the analyses, ensuring that the results were both qualitatively and quantitatively similar. The variables that follow normality had a *t*-test performed on them, and the variables that were non-normal had the nonparametric test (Mann-Whitney *U*-test) performed on them.

At the multivariate level, linear discriminant analysis was used to assess the performance of the BSA. This statistical approach was used to formulate a function or rule(s) that would permit subjects to be classified into one of the following predefined classifications: (1) individuals with osteoporosis who had 1 or more vertebral fractures and (2) individuals with osteoporosis without vertebral fractures. The linear discriminant analysis follows closely the logic of the multiple regression analysis. The linear discriminant analysis was performed on those variables that showed a significant difference in means between the 2 groups (fractured and nonfractured osteoporosis patients) at the 0.05 significance level. Age was also considered for the linear discriminant analysis. The performance of the classification rule was summarized by means of misclassification error. To obtain a reliable estimate of future performance, one should use a data set other than that used to derive the classification rule. However, in the face of limited resources with small sample size, such as that used in this study, the data were repeatedly split into 2 parts, creating the rule on 1 data set and testing the rule on the remaining data set. The method of cross-validation (leave one out) was used. Here, a single subject was chosen for testing, and the rest of the data were used to create the discriminant function. This was repeated for all the subjects. The predicted error rate of the rule based on all the subjects was estimated as the proportion of misclassifications. Receiver operating characteristics (ROC) curves were also constructed for each of the variables under BSA to classify the nonfracture and fracture groups separately for each of the leg. Furthermore, using ROC analysis, area under the curve (AUC) and cutoff of the damping variables were also obtained to assess the performance of the variables separately for each of the leg. The cutoff was decided where both sensitivity and specificity were found to be higher. In addition, ROC analysis was also carried out for each of the BMD variables.

Results

Table 1 provides demographic data, dietary calcium intake, and fracture status of the study groups. The age range of patients in the nonfracture group was 65−86 yr and, in the fracture group, it was 67−85. There was no statistically significant difference between fracture and nonfracture groups with regard to body weight, height, BMI, or dietary calcium intake. Age was significantly different between the groups ($p = 0.0133$).

BMD: Table 2 provides mean values of BMD and T-scores of femoral neck, trochanter, total hip, and lumbar spine for both groups. There was no statistically significant difference between the fracture and the nonfracture group for any of the BMD outcomes.

Static bone quality property: Table 3 provides biomechanical variables calculated from femoral neck geometry. None of the biomechanical variables were significantly different between the fracture and nonfracture groups.

Dynamic bone quality measure—BSA: Figures 2 and 3 provide bone damping values (ζ) of the right and left leg testing along with values from healthy young adults *(25,26)*. Based on the results of right-foot heel strike, the fracture group showed statistically significant lower ζ than the nonfracture group in all anatomical sites (range of p values was 0.033−0.001) (Fig. 2). However, for the left foot heel strike, the fracture group showed statistically significant lower ζ than the nonfracture group for all anatomical sites except for the below knee site (range of p values was 0.047−0.001) (Fig. 3). As expected, the ζ values at the fracture region (i.e., between upper [$\zeta_{\text{Upper back}}$] and lower back [$\zeta_{\text{Lower back}}$]) were much lower than those recorded for the remaining sites of both groups (Figs. 2 and 3). Compared with the nonfracture group, on average, across all anatomical sites, the fracture group showed 40% and 50% lower damping for the left and the right heel strikes, respectively. Overall, the resonance frequencies at all anatomical sites were numerically higher for the fracture group than the nonfracture group, but the differences were not statistically significant. At the spine, the average resonance frequency in the fracture group was 47−52% higher than that of nonfracture group for the lower back and the upper back (figures not shown).

The results from linear discriminant analysis performed on those variables that showed a significant difference ($p \leq 0.05$) in mean values between the 2 groups (fractured and

Table 1
Demographics and Dietary Calcium Intake

Variable	Fracture	Nonfracture
Body weight (kg)	59.4 ± 1.6	59.6 ± 1.5
Body mass index (kg/m^2)	23.3 ± 0.6	23.3 ± 0.6
Age (yr)	75.8 ± 1.0*	72.2 ± 0.9
Height (cm)	159.7 ± 1.3	158.9 ± 1.1
Calcium intake (mg/d)	1880 ± 127	1690 ± 101

*$p = 0.013$.
Results are given as mean ± standard error of the mean.

Table 2
BMD and T-Scores

Anatomic Sites	Fracture	Nonfracture
Femoral neck BMD (g/cm^2)	0.570 ± 0.018	0.590 ± 0.012
T-score	−2.6 ± 0.17	−2.4 ± 0.11
Trochanter BMD (g/cm^2)	0.469 ± 0.018	0.495 ± 0.013
T-score	−2.6 ± 0.20	−2.3 ± 0.14
Total hip BMD (g/cm^2)	0.623 ± 0.014	0.646 ± 0.014
T-score	−2.8 ± 0.18	−2.7 ± 0.13
Lumbar spine BMD (g/cm^2)	0.678 ± 0.023	0.726 ± 0.016
T-score	−3.3 ± 0.21	−2.9 ± 0.14

GE-Lunar results were converted to Hologic units, and T-scores for GE-Lunar hip sites were calculated from the converted units. There was no statistically significant difference between groups.
Abbr: BMD, bone mineral density.

Table 3
Biomechanical Variables From Femoral Neck Geometry
(Hologic Only; Fracture = 16; Nonfracture = 18)

Biomechanical Variables	Fracture	Nonfracture
Compressive strength (g/cm)	1.96 ± 0.10	1.98 ± 0.07
Buckling ratio (unitless)	17.6 ± 1.2	16.2 ± 0.4
Section modulus (cm^3)	0.99 ± 0.06	0.98 ± 0.04
Cortical thickness (cm)	0.105 ± 0.003	0.108 ± 0.003

There was no statistically significant difference between groups.

nonfractured) for FRF analyses provided the following model for left leg heel strike case:

$$\text{Group (fracture or no fracture)} = 4.4 + 0.188\,\zeta_{\text{Lower back}} + 0.817\zeta_{\text{Upper back}} - 0.086\,\text{Age}$$

Using the aforementioned discriminate analysis for left leg, about 74% of original grouped cases were correctly classified. In the aforementioned equation, negative coefficient for age implies that older patients are more likely to be in fracture group; positive coefficients for $\zeta_{\text{Lower back}}$ and $\zeta_{\text{Upper back}}$ imply that individuals with higher damping at the lower back and upper back are less likely to be in the fracture group.

For the right leg heel strike case, the following model was obtained:

$$\text{Group (fracture or no fracture)} = -1.283 + 0.147\,\zeta_{\text{Above knee}}$$

Using the aforementioned discriminate analysis for right leg, about 66% of original grouped cases were correctly classified. In the aforementioned equation, positive coefficient for $\zeta_{\text{Above knee}}$, as expected, implies that a higher damping at right above knee is associated with less likelihood of being in the fracture group.

In Figs. 4 and 5, results from ROC analysis are provided for right and left leg testing. In these figures, ROC analysis

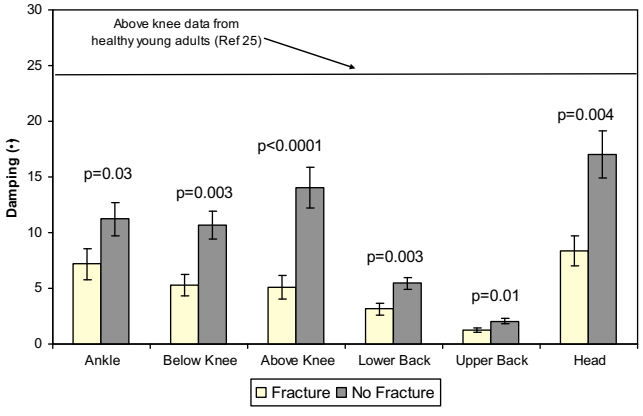

Fig. 2. Mean ± standard error of the mean. Damping values at various anatomical sites of patients with osteoporosis (ankle, below knee, and above knee are for right leg).

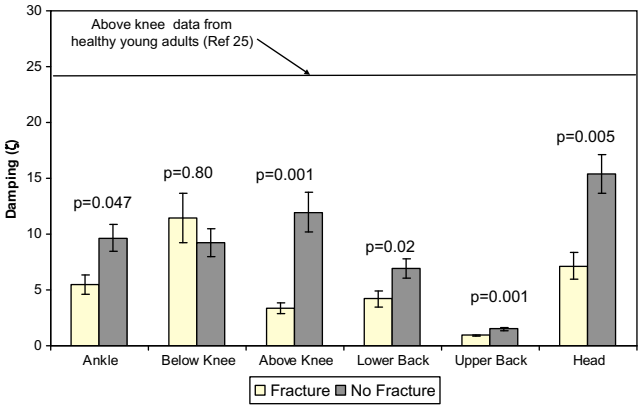

Fig. 3. Mean ± standard error of the mean damping values at various anatomical sites of patients with osteoporosis (ankle, below knee, and above knee shown for left leg).

provides relative comparison of AUC values for damping (ζ) measured at various anatomical sites. For the left leg, $\zeta_{\text{Lower back}}$ was the best discriminator (AUC = 0.91 [95% confidence interval (CI): 0.83−1.0]), with $\zeta_{\text{Above knee}}$ (AUC = 0.87 [95% CI: 0.76−0.98]) being the second best. The remaining anatomical sites at head, ankle, upper back, and below knee had AUC values of 0.82 (95% CI: 0.68−0.96), 0.75 (95% CI: 0.59−0.92), 0.72 (95% CI: 0.56−0.88), and 0.82 (95% CI: 0.68−0.97), respectively. For the left leg test, the cutoff values for ζ at head, above knee, ankle, lower back, below knee, and upper back were 9.0, 3.87, 5.55, 4.33, 6.12, and 1.04, respectively.

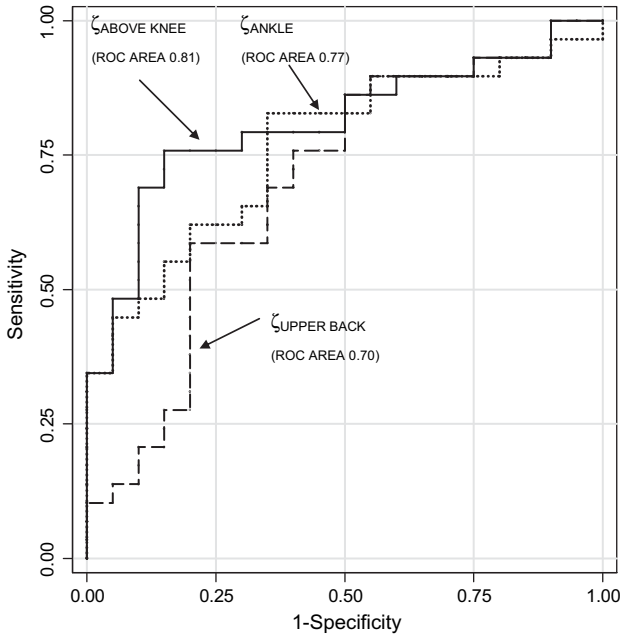

Fig. 4. Receiver operating curves for damping (ζ) at 3 anatomical sites showing the top 2 and the lowest values of area under the curve (AUC) for right leg testing. The remaining anatomical sites at head, lower back, and below knee had AUC values of 0.75, 0.76, and 0.74, respectively.

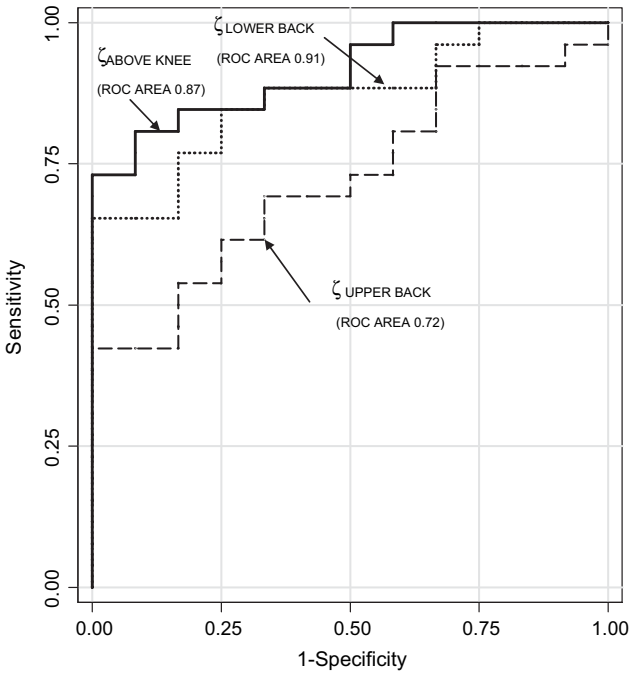

Fig. 5. Receiver operating curves for damping (ζ) at 3 anatomical sites showing the top 2 and the lowest values of area under the curve (AUC) for left leg testing. The remaining anatomical sites at head, ankle, and below knee had AUC values of 0.82, 0.75, and 0.82, respectively.

For the right leg, the best discriminator was $\zeta_{\text{Above knee}}$ (AUC = 0.81 [95% CI: 0.67−0.93]) and the second best was ζ_{Ankle} (AUC = 0.77 [95% CI: 0.63−0.9]). The remaining anatomical sites at head, lower back, upper back, and below knee had AUC values of 0.75 (95% CI: 0.61−0.89), 0.76 (95% CI: 0.62−0.90), 0.70 (95% CI: 0.54−0.85), and 0.74 (95% CI: 0.61−0.88), respectively. For the right leg test, the cutoff values for ζ at head, above knee, ankle, lower back, below knee, and upper back were 9.66, 5.95, 7.04, 3.21, 5.32, and 1.15, respectively.

The ROC analysis of the femoral neck, trochanter, total hip, and lumbar spine BMDs provided the following AUC values: 0.52 (95% CI: 0.35−0.69), 0.55 (95% CI: 0.38−0.72), 0.57 (95% CI: 0.40−0.73), and 0.63 (95% CI: 0.48−0.78), respectively.

Discussion

We describe a noninvasive measure of bone quality, BSA property, damping factor (ζ), as an aggregate response of bone's structural integrity under dynamic "realistic" in vivo loading associated with simple heel strike. In women with osteoporosis, the BSA property, ζ, was able to discriminate between those with and without fracture, whereas BMD and static biomechanical measures alone were not significantly different between the groups, as would be expected from the literature *(38−40)*.

Based on the literature, skeletal disorders, such as osteoporosis, detrimentally impact the bone's quality, characterized by decreases in section modulus, cortical thickness, and compressive strength, and increase in buckling ratio *(38)*. These traditional measures of bone quality, static biomechanical properties, based on bone geometry data (from DXA testing), are estimated using classical strength of materials equations from the literature *(29)*. We calculated these variables in the subset of our patients using the method and software published by Riancho et al and compared those with data from the literature *(29,37)*. Although our fracture group showed lower values for section modulus, compressive strength, and cortical thickness and higher values for buckling ratio than the nonfracture group, these differences were not statistically significant (Table 3). Compared with the values of static biomechanical properties of healthy subjects reported in the literature, our subjects had much lower values, suggesting poorer structural integrity of their bones *(29,38)*.

As discussed earlier, bone quantity, as measured by BMD, and traditional measures of "bone quality," described by static biomechanical properties, were not sensitive enough to discriminate between fracture and nonfracture groups of patients with osteoporosis. Fracture is a structural failure phenomenon, and structural failure of human bone and/or any mechanical system rarely occurs under static conditions *(12)*. Therefore, better understanding of bone fracture and prevention requires that traditional measures of BMD and static biomechanical properties be supplemented with measures of dynamic biomechanical properties of bone when exposed to realistic "real world" in vivo external loadings. The BSA measurement was designed to capture "dynamic bone quality" properties (as measured by BSA outcome, damping factor ζ) noninvasively when exposed to realistic dynamic loading associated with simple heel strike. Because osteoporosis is associated with decreased bone mass and deterioration of trabecular architecture that collectively impact bone's mechanical properties, it will detrimentally change bone's natural shock-absorbing capacity, which can be quantified with the BSA measurement system. In our current study, BSA outcome, ζ, was capable of discriminating between postmenopausal osteoporosis patients with and without vertebral fractures. Although the fracture was limited to the spinal region, ζ was also significantly different between the groups at all anatomical measurement sites, implying that structural integrity was compromised systemically (Figs. 2 and 3). Based on these results, it appears that decreased ζ observed at the site of fragility fracture was also present at other load-bearing anatomical sites, such as tibia and femur. However, as expected, the absolute ζ values (compared with healthy young adults [n = 10], mean ζ = 24.3 [standard error of the mean (SEM): 1.9]), because of right leg heel strike at the fracture sites ($\zeta_{\text{Upper back}}$: 1.26 and $\zeta_{\text{Lower back}}$: 3.15), were significantly lower than those at tibial (ζ = 5.3) and femoral (ζ = 5.1) bone sites, because osteoporosis will have greater detrimental impact on damping properties of the bone at the fracture site compared with nonfracture anatomical sites (Fig. 2) *(25,26)*.

Based on the ROC analysis, BSA outcomes at all anatomical sites provided much higher AUC values than those obtained for the BMD outcomes. The ranges of AUC values for the BSA outcomes for the left and the right leg tests were 0.72−0.91 and 0.7−0.81, respectively. On the other hand, range of AUC values for the BMD outcomes was 0.52−0.63. Compared with AUC for BMD, the BSA outcomes have better capability in discriminating patients with osteoporosis with and without fracture.

The linear discriminate analysis models of the left and the right leg tests show that sensitive BSA variables were $\zeta_{\text{Lower back}}$ and $\zeta_{\text{Upper back}}$ and $\zeta_{\text{Above knee,}}$ respectively, for correctly identifying fracture category. It is not clear why 2 different sets of BSA variables were found to be associated with left and right leg tests, as there were no experimental differences in carrying out the left/right leg testing. The mean values of damping factor, ζ, for the fracture and the nonfracture groups were about 78% and 42% lower, respectively, than those reported for young healthy adults and youths *(25,26,37)*. Although such a drastically lower damping value in osteoporosis compared with a normal healthy bone is not surprising, it is not clear as to what extent normal aging affects the damping properties. Further studies are needed to address the age-associated issues of damping properties.

In the present study, calculation of ζ was based on classical structural dynamic theory, which directly relates damping of a structure to its viscous damping coefficient and inversely to the square root of its resonance frequency *(34−37)*. Therefore, a lower value of ζ and a higher resonance frequency found in the patients with osteoporosis with and without fracture compared with healthy adults implies lower viscous damping properties of the bone among the patient groups. Such a finding is consistent with a recent report of collagen loss as a contributing factor in decreased viscoelastic properties of bones and potential association of osteoporosis with pathogenesis *(41,42)*. Previous studies with bone turnover markers have also suggested that bone matrix component type I collagen level is a contributing factor in determining damping/viscoelastic capacity of bone *(41−44)*. Therefore, in our study, a lower value of ζ of the fracture group implies that their bones have lower viscous damping capacity and, therefore, are stiffer than that of the nonfracture group.

We believe that ''dynamic bone quality'' property damping factor, ζ, is an aggregate response of bone's structural integrity under ''realistic'' in vivo loading. The bone's response under realistic dynamic loading provides a better picture of its structural integrity than that obtained under unloaded condition, as structural failure of human bone and/or any mechanical system rarely occurs under static conditions *(12,13)*. Recent studies indicate that static testing alone does not provide true response of the composite materials constituting the bone *(13)*. As bone is a composite material comprised of minerals and organic materials as well as water, its viscoelastic properties describing its damping capacity can only be truly captured when the bone is exposed to realistic self-induced in vivo loading (such as heel strike used in our study) that permits aggregate response as a result of interactions of some or all of these components of bone *(13,45)*. This comparison between measures of dynamic bone quality and static bone quality can be analogous to comparing the evaluation of heart conditions using a treadmill-based, stress-induced electrocardiogram (ECG) (dynamic test) with that by a resting ECG (static test). It is well known that a stress-induced ECG reveals more about the heart condition than a resting ECG.

Although we measured damping capacity of bone by in vivo mechanical stimulation associated with self-induced heel strike in our study, others have estimated damping properties using externally applied acoustic energy in an animal model *(46,47)*. These animal studies have shown that measures of damping properties are an early indicator of bone's structural integrity compared with conventional tools. Similarly, the BSA technique used in our study measured damping factor ζ, which is better at discriminating between osteoporosis patients with and without fracture than the conventional measures of BMD and static biomechanical properties *(26,37)*. Obviously, future studies are needed to evaluate the ''dynamic bone quality'' measure, ζ, as a sensitive measure of a bone's structural integrity for predicting fracture and discriminating patients with varying degrees of severity of osteoporosis with and without fracture. The results presented in our study and those in the literature suggest that precise prediction of fracture risk should combine both measures of static and dynamic mechanical properties as well as BMD *(13,48)*.

The limitations of our study include a small sample size; therefore, findings should be interpreted with caution. Second, attachment of skin-mounted accelerometers might have introduced some relative movement between the bone and the skin, but this was minimal, as we used low-mass accelerometers with appropriate band pass filters that have been found to be effective in accurately representing bone vibration by others *(30,31)*. Because the accelerometers were attached on the bony prominences of the whole bone, the response obtained was reflective of aggregate damping value for both trabecular and cortical bone. Biomechanical properties, calculated using bone geometry obtained from DXA (measured by Hologic only), have limitations, because shapes were assumed to be symmetrical, which might not be true in all directions *(48)*. Although Lunar-based DXA data could have also been used for the calculation of biomechanical properties, it assumes that both devices are calibrated identically to mineral mass, which might not be necessarily true. With such assumptions in calibration, it is possible that biomechanical values obtained by these formulae would be crude estimates.

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