Development of Optical Bone Densitometry Using Near-infrared Light

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ABSTRACT

In order to realize the early detection of osteoporosis, development of a device that enables simple screening is desirable. We have focused on a simple bone density evaluation method using near-infrared light and have studied its reliability. In this study, we propose an optical bone density measurement method, in which bone density is evaluated by the slope of light intensity on the skin surface. A detection system of quasi-ballistic light is incorporated into this method for better accuracy, by decreasing the influence of light scattering in the skin layer. This method was verified by Monte Carlo simulation and experiments using bone and skin phantoms. In the simulation, we made an analytical model, based on the proposed method, to investigate the propagation behavior of photons in skin and bone tissues, and the light intensity distribution on the surface of the skin was analyzed. In the experiment, a computer-controlled apparatus was developed, based on the proposed method, for the detection of light intensity distribution on the surface of the skin phantom and calcined bovine cancellous bone, respectively. The skin surface was irradiated with near-infrared laser light (wavelength: 850 nm) and the light intensity was measured over the surface using the apparatus. A positive correlation was obtained for the relationship

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between the slope of the light intensity distribution and the bone density, within the range of 2 mm or less of skin thickness. This experimental result was qualitatively supported by the simulation results. In particular, the correlation in the range of 1 to 1.5 mm of skin thickness suggested an acceptable accuracy for this method.

Keywords: *Near-infrared light, Bone density, Non-invasive measurement, Osteoporosis.*

Introduction

X-ray is currently used for osteoporosis diagnosis, including dual energy X-ray absorptiometry (DXA) [1] and quantitative computed tomography (QCT) [2]. However, it is difficult to utilize these devices for the early detection of osteoporosis, which only has a moderate decrease in bone mineral density (BMD), due to their large size and potential harmfulness, indicating the necessity for a compact and safe device for screening of osteoporosis. Near-infrared light is useful for non-invasive measurement of different types of bio-information and is utilized in the medical field. However, its application to detection of bone density has not been studied or developed sufficiently, because of the difficulty in eliminating the scattering effect of the skin layers over bone tissue. We propose a novel optical bone densitometry method that decreases the effect of optical scattering in the skin by detection of quasi-ballistic light. The purpose of this study was to validate this proposed method by using Monte Carlo simulation and by *in vivo* experiments using bone and skin phantoms.

Materials and methods

Principle of optical bone densitometry

When the skin surface over bone tissue is irradiated with laser light, light intensity distribution is observed on the outer surface of the skin and the slope of the distribution should correlate with bone density. At high bone density, the light intensity abruptly decreases with an increase in distance from the light irradiation position. In contrast, at low bone density, a moderate intensity reduction appears. Therefore, bone density can be evaluated from the slope of the light intensity distribution. In our proposed method, by using an optical system composed of a laser diode, two round slits, and two convex lenses, as shown Fig. 1, quasi-ballistic light from the skin surface is selectively detected by a photodiode. By moving this optical system toward the object tissue (Z direction), the light intensity is continuously acquired over the distance from the irradiation point on the skin surface. In addition, since the skin has lower light scattering properties compared with bone, light tends to go straight. Thus, the slope of the

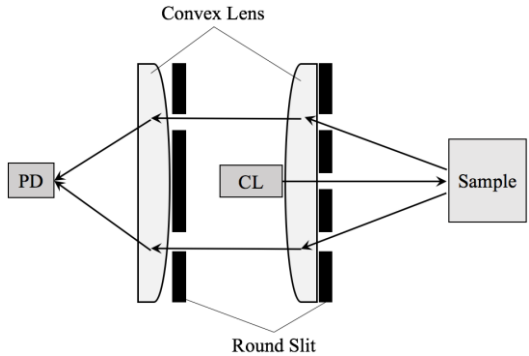

Fig.1 Composition of the optical bone densitometer.CL: coherent light, PD: Photo diode

intensity distribution shows almost the same value for the same bone density, regardless of the difference in skin thickness, if quasi-ballistic light is detected. This principle suggests that our method is robust regardless of skin variation, giving better accuracy for the prediction of the bone tissue density under the skin layer.

Monte Carlo simulation

The proposed optical bone densitometry method was confirmed by numerical simulation using the Monte Carlo method. The Monte Carlo method is performed by repeating numerical calculations using random numbers, and gives us the approximate propagation behavior of photons in a substance. Propagation distance L, photon energy decrease amount ΔW, and scattering direction θ of photons in a substance, are calculated as follows: [3][4]

$$L = \frac{-ln(\xi)}{\mu_a+\mu_s} \tag{1}$$

$$\Delta W = \frac{\mu_a}{\mu_a+\mu_s} W \tag{2}$$

$$\theta = cos^{-1}\begin{cases}\frac{1}{2g}\left\{1+g^2-\left(\frac{1-g^2}{1-g+2g\xi}\right)^2\right\} & if\ g > 0 \\ 2\xi - 1 & if\ g = 0\end{cases} \tag{3}$$

Here, μ_a and μ_s and g are absorption, scattering, and anisotropy coefficients, respectively. Anisotropy coefficient g takes a value from 0 to 1, indicating complete isotropic scattering when g = 0. In addition, ξ is a random number taking a value from 0 to 1. In this study, an optical model was created based on the principle of the proposed method, and programmed in Python 3.5, adopting optical properties for skin and bone tissues shown in Table 1[5] [6].

Table 1 Optical properties and the thickness of skin and bone layers [5] [6]. μ_a: absorption coefficient. μ_s: scattering coefficient. g: anisotropy of scattering. d: thickness.

Sample	μ_a (cm^{-1})	μ_s (cm^{-1})	g	d (cm)
Skin	0.5	150	0.9	0.05~0.2
Bone	0.2	50~400	0.9	100

Bone densities are expressed by scattering coefficients in this simulation, since bone density positively correlated with the scattering coefficient [6]. In the optical path analysis in the optical system, the size of circular slits (average diameter 20 mm, gap 2 mm, thickness 3 mm), lenses, and the detection surface of the photodiode (PD, ø3 mm) were determined by referring to actual commercial products. Light intensity is expressed as absorbance (Abs), which is calculated from the number and weight of photons reaching the photodiode, as follows:

$$\mathrm{Abs} = \ln\left(\frac{\sum_{i=1}^{n_d} W(i)}{n_t}\right) \quad (4)$$

Here, *W(i)* represents the weight of each photon, and n_t is the total number of photons detected by the photodiode. The total number of incident photons in one simulation was 10^7.

In vivo experiments using tissue phantoms

Fig. 3 shows the internal configuration and external appearance of an experimental device developed for the validation of the proposed method. This gun-type compact device is composed of circular slits, lenses (48795 and 48797, Edmund Optics Japan Co., Ltd), the laser diode (LD, wavelength: 850 nm, LDM115G/850/1, Grobal Laser Ltd, UK), and the PD (54522-h, Edmund Optics Japan Co., Ltd), packed into the optical unit as shown in Fig. 1. The optical unit is moved in the Z direction by a stepping motor driven linear actuator (LX2001P-B1-T2028-80, Misumi Co., Ltd, Japan). A microcomputer (Arduino Uno, Arduino SRL, Italy) controls the motor and collects signals from a photo-amplifier (C9051, Hamamatsu Photonics K.K., Japan). Travel distance of the optical unit is calculated by the number of pulse signals sent to the stepping motor. Light intensity data are sent to the laptop (MacBook Air 6,2, Apple Inc., America) by serial communication, and converted to absorbance (Abs) as follows:

$$Abs = ln\left(\frac{I}{I_0}\right) \quad (5)$$

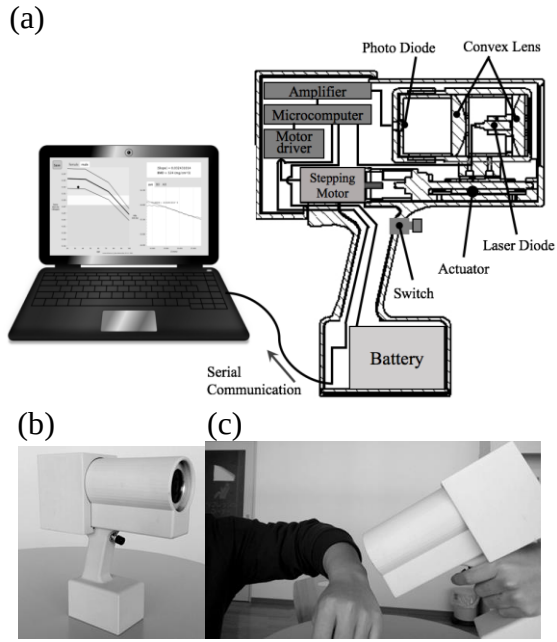

Fig. 3 Internal configuration (a) and external appearance (b, c) of the optical bone densitometer.

Here, I_0 is the incident light intensity from LD, and I is the detected light intensity. Finally, from the relationship between the travel distance and the absorbance, i.e., the intensity distribution, the slopes are determined by a regression line for the selected region of the data and compared with the densities of bone phantoms.

Calcined bovine cancellous bone and compact bone were used as bone phantoms. Cancellous bone cubes with one side of about 4 cm were obtained from a bovine femur, boiled to remove the bone marrow, and calcined at 600 °C for 24 hours in an electric furnace. Nine calcined cancellous bones were prepared for the experiments. Bovine cancellous bone is not homogeneous; therefore, different densities were observed in the same cancellous bone cube. To clarify the density of the measurement region, the bone phantoms were scanned by μCT (InspeXio SMX-90CT Plus, Shimadzu Corporation, Japan). A cubic region of 1 cm per side was selected at three different positions for each bone phantom, and the bone density at each position was analyzed by the bone morphometry software (TRI/3D-BON-FCS64, Ratoc System Engineering CO., Ltd., Japan). Compact bone plate with a thickness of 0.3 mm was obtained by slicing the bovine femoral

diaphysis in parallel to the bone axis, and calcined in the same manner as for the cancellous bones. The thickness of the compact bone phantom was adopted from the cortical bone thickness at the human distal radius, which is the measurement site for our bone densitometry (Fig. 3(c)), by referring to the sectional image database of the Visible Human FTP Resource [7], provided by the United States National Library of Medicine (NLM). The concentration of intralipid (Intralipos®, Otsuka Pharmaceutical Co., Ltd, Japan) was adjusted to 2% with water to reproduce the optical properties of skin [5]. The 2% intralipid was enclosed between two cover glasses by a shim ring and used as a skin phantom. By changing the thickness of the shim ring, skin phantoms with different thicknesses from 0.5 to 2 mm were prepared.

Results and discussion

Fig. 4 (a) shows the intensity distribution obtained by the Monte Carlo simulation in the case of a 1 mm skin thickness. Intensities, which are represented by absorbance $\ln(I/I_0)$, were decreased with the increase in Z in all the scattering coefficients. Fig. 4 (b) shows the relationship between the slopes in the range of Z = 30 to 37 mm and scattering coefficients, demonstrating a positive correlation ($r^2 = 0.60967$), however, larger variations remained at the lowest and highest scattering coefficients.

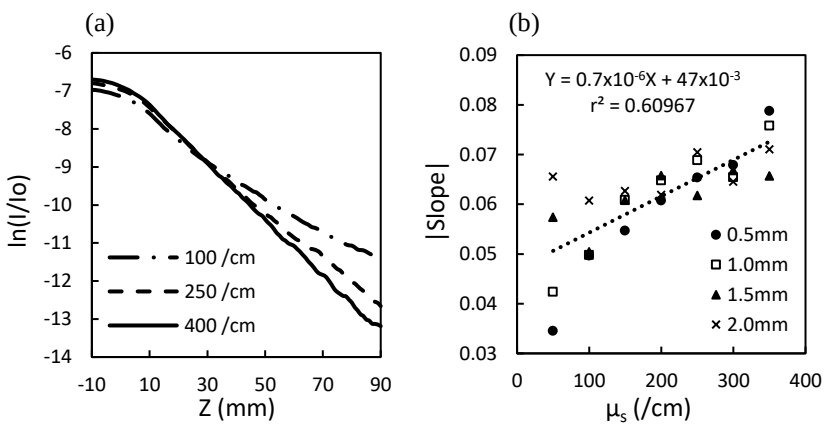

Fig. 4 (a) Intensity distributions simulated by the Monte Carlo method for each different scattering coefficient. Z is the distance between the phantom surface and the focal point of the lens with positive values to the depth direction. (b) Relationships between scattering coefficient, representing bone density, and the slope of intensity distribution in different thicknesses of skin.

On the other hand, in skin thicknesses of 0.5 to 1.5 mm, the slope value was relatively close at any scattering coefficient, suggesting that stable measurement against the skin effect is possible within this range of skin thickness.

Fig. 5 (a) shows the intensity distributions for three representative densities of bone phantoms obtained experimentally, with a 1 mm thick skin phantom on the bone. Intensity was attenuated with the increase of Z, similar to the results demonstrated by the Monte Carlo simulation. Fig. 5 (b) shows relationships between the density of the bone phantom and the slope of intensity distribution with the different thicknesses of skin. Each slope was determined by line regression in the range of Z = 30 to 37 mm. The relationship between the density and the slope demonstrated a good positive correlation, although separately in skin thickness of 1.0 and 1.5 mm, and 0.5 and 2.0 mm. In the former range, the sensitivity of the slope to bone density was relatively higher than that in the latter range.

The above results from the Monte Carlo simulation and the *in vivo* phantom experiments showed a qualitative agreement, suggesting the feasibility of our proposed optical bone densitometry, however only applicable to the limited range of skin thicknesses from 1.0 to 1.5 mm. To expand the valid range of this method and improve the accuracy, other parameters, not only slope, should be added to the database of optical intensity distributions.

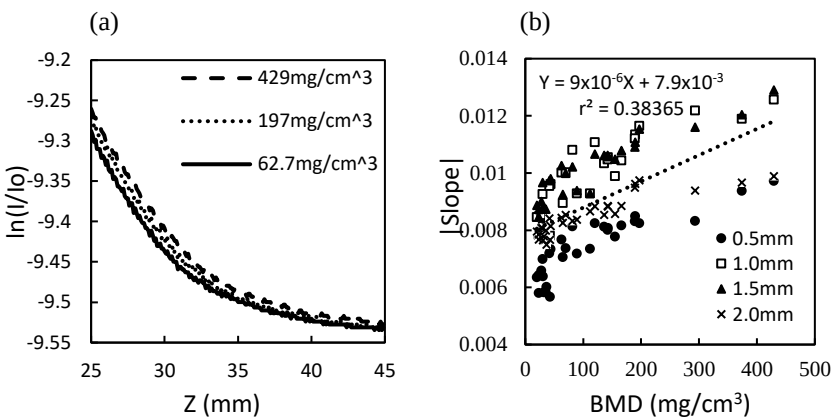

Fig. 5 (a) Intensity distributions of three bone phantoms of different densities with a skin phantom of 1 mm thickness. (b) Relationships between the density of the bone phantom (BMD) and the slope of intensity distribution in the different thicknesses of skin.

The slopes obtained from the Monte Carlo simulation were larger than those from the *in vivo* experiment. This inconsistency seems to be derived from the internal bone microstructure (trabecular structure) in cancellous bone. Despite the microstructure of the bone, in the Monte Carlo simulation, bone tissue was treated as a solid layer. Therefore, the simulation model should be improved by adopting the 3D-structural data of cancellous bone from micro CT for better results.

Conclusion

In this study, a bone density measurement device using near-infrared light was proposed and validated by Monte Carlo simulation and *in vivo* phantom experiments. From the results obtained, it is concluded that this method is applicable to non-invasive prediction of BMD with limited skin thicknesses, suggesting possible availability for screening of osteoporosis.

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