First assessment of three-dimensional quantitative photoacoustic tomography for in vivo detection of osteoarthritis in the finger joints

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Purpose: The purpose of this pilot clinical study is to assess three-dimensional (3-D) quantitative photoacoustic tomography (qPAT) for in vivo detection of osteoarthritis (OA) in the finger joints.

Methods: All subject data were handled in compliance with the rules and regulations concerning the privacy and security of protected health information under HIPAA. Seven female subjects (two OA patients and five healthy controls) entered the study and their distal interphalangeal (DIP) joints were examined by a 3-D photoacoustic scanner. 3-D optical absorption coefficient images of all the photoacoustically examined joints were recovered using a 3-D qPAT reconstruction algorithm.

Results: The recovered quantitative photoacoustic images revealed obvious difference in the optical absorption coefficient of the joint cavity (cartilage and synovial fluid) between the OA and healthy joints. Quantitative analysis of the joints also indicated an apparent difference in the recovered joint spacing between the OA and healthy subjects.

Conclusion: This initial clinical evaluation suggests that it is feasible to detect osteoarthritis in the finger joints with our 3-D qPAT approach, which has paved the way to further statistically evaluate the diagnostic performance of the 3-D qPAT approach in comparison with radiography or magnetic resonance imaging (MRI) on a sample of hand osteoarthritis. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3598113]

I. INTRODUCTION

Osteoarthritis (OA) is the most prevalent degenerative joint disease, which involves tens of millions of Americans with disability or diminished quality of life. Primary OA affects not only large weight-bearing joints in the hips and knees but also small joints in the hand.^{1,2} Although knee and hip OA bear most responsibility for the burden of OA, hand OA may be a marker of a systemic predisposition toward OA, among which the second distal interphalangeal joint is most frequently affected (57% in women and 37% in men) from an incidence study.^{2,3} The typical pathological features of OA include erosion of articular cartilage, bony changes, and occasional synovial effusion as well as associated enhanced blood vessel growth.^{3,4}

Imaging techniques, including x-ray radiography, computed tomography (CT), ultrasonography, and magnetic resonance imaging (MRI), have been conventionally used or investigated to detect the physiologic/anatomic abnormalities in the joint cavity when OA has been established.^{5–17} However, clinical examination remains the principal approach while these available imaging techniques are used only as supplemental methods in OA diagnosis, especially in hand OA, either because they are insensitive to the abnormal changes in the OA joint cavity or because they are too costly to be used as a routine examination method. For example, plain radiographs are conventionally used to visualize bony abnormality; however, they are insensitive to changes in soft tissues (cartilage, synovial fluid, etc.) and therefore, incapable of capturing the primary features of early stage OA.⁷ As such, sensitive and affordable imaging methods are urgently needed for the detection of OA, especially in an early stage. Moreover, the progress of effective imaging methods in OA

diagnosis may at the same time accelerate the advancement of medical therapies for OA, which are currently used to relieve OA symptoms or to prevent the worsening of OA to certain extent with limited effectiveness in modifying OA. For example, sensitive imaging techniques can possibly serve as surrogate markers of clinically meaningful outcomes to economically and efficiently validate the candidate disease-modifying OA drugs.³

Near-infrared (NIR) light based imaging techniques have been recently studied to image finger joints and to detect joint diseases at an early stage by visualizing highly sensitive optical absorption/scattering abnormalities in soft tissues (cartilage, synovial fluid, etc.) that are absent from other imaging modalities.^{18–25} While pure-optical imaging techniques, such as diffuse optical tomography, are able to detect the optical absorption/scattering abnormalities associated with soft tissues in OA and RA joints, its spatial resolution is relatively low (about 3–5 mm) for imaging tissues in approximately centimeter depth.^{18–21} Photoacoustic tomography (PAT), which combines both the high optical absorption contrast and high ultrasound resolution, is able to visualize the same sensitive “color” abnormality (optical absorption contrast) with significantly improved spatial resolution (0.5 mm or better, adjustable with ultrasound frequency) for deep-tissue imaging in optically diffusive regime. For example, tendon, aponeurosis, volar plate, subcutaneous tissue, phalanx, and other tissues around DIP joints were well differentiated when a cadaver human finger was photoacoustically scanned in cross section with a 10 MHz transducer.²² Up to now, PAT has shown its potential to detect breast cancer, to assess vascular and skin diseases, to monitor epilepsy in small animals, to image finger joints, to visualize

fluorescent proteins, and to evaluate exogenous contrast agents in molecular imaging.^{22,23,26–28} By combining a light transport model with the conventional PAT method, quantitative PAT (qPAT) is able to provide quantitative tissue absorption coefficient and physiological/functional parameters including hemoglobin concentration, blood oxygenation, and water content when multispectral light is used, which can further quantify the neoangiogenesis and hypoxia in diseased finger joints for more accurate diagnosis and better understanding of OA development.^{39–43}

The goal of this paper is to present a pilot clinical study that aims to assess the potential of 3-D qPAT as a sensitive detection method of OA in the finger joints.

II. MATERIALS AND METHODS

II.A. Three-dimensional quantitative photoacoustic imaging

Our 3-D qPAT system [Fig. 1(a)] consists of a pulsed NIR light source, a spherical scanning subsystem, an ultrasound detection array, and associated data acquisition system. In this system, pulsed light beam at 805 nm is generated by a Ti: Sapphire laser (LOTIS, Minsk, Belarus) with a repetition rate of 10 Hz and a pulse width of <10 ns, and uniformly illuminates the DIP joint of interest. The light beam is guided via an optical fiber and the laser energy at the DIP joint is controlled around 10 mJ/cm^2 , which is far below the safety standard of 22 mJ/cm^2 .⁴⁴ The ultrasound detection array is composed of eight 1 MHz circular flat transducers (Valpey Fisher, Hopkinton, MA) arranged equally spaced along a 210° arc arm, which allow for 240° detection along the circle maximally covering the distal end of the finger joints. Each of the 1 MHz transducers has a bandwidth of 80% at -6 dB and an element size of 6 mm, which was chosen to optimize the imaging of joint tissues and bone structures. The position and performance of each transducer in the ultrasound array was calibrated carefully with controlled tissue phantom experiments. During an exam, the palmar side of the DIP joint faced up, allows the DIP joint to be illuminated from the dorsal side of the finger. Our experience indicates that this way of light illumination can give us maximized tissue penetration. To secure the position of the finger, rubber bands are applied to the distal

tip and the proximal end of the finger. Both the ultrasound array and the examined finger are immersed in a water tank for minimized ultrasound attenuation. The detected acoustic signals are fed into preamplifiers and converted to digital signals by an A/D board (PREAMP2-D and PCIAD1650, US Ultratek, Concord, CA). LABVIEW programming is used to control the entire examination procedure. The acoustic signals at 360 scanning locations over a spherical detecting surface, surrounding the DIP joint of interest, were collected through the combined rotations of two rotary stages with orthogonal rotary axes. The scanning procedures were described in detail previously.²³ In brief, one rotary stage controls the ultrasound detecting array to scan 40 circular locations with an interval of 6° in the plane of ultrasound array, covering a total of 240° along the circle surrounding the distal end of the finger joints, while the other rotary stage tilts the ultrasound detecting array from -60° to 60° with the horizontal plane by a step of 15° , allowing for nine detecting angles and a 240° coverage along the cross section of the finger joint. With the ultrasound array, the whole examination takes 5 min.

The optical absorption coefficient images of the DIP joints are quantitatively reconstructed by using a finite element mesh of 7519 nodes and 41323 tetrahedral elements with a 3-D qPAT reconstruction algorithm described elsewhere.^{40,41} In brief, the photon density can be calculated from the 3-D photon diffusion equation, and the estimated optical absorption coefficients are iteratively updated by dividing the absorbed optical energy reconstructed in conventional PAT by the calculated photon density, until the product of the calculated photon density and the current estimated optical absorption coefficients is close enough to the absorbed optical energy reconstructed. A full-width at 30% maximum (FW30%M) method adapted from optical imaging techniques is used to quantitatively analyze the absorption coefficient profiles from the recovered 3-D images, which is capable of improving the measurement accuracy in determining the thickness of optically imaged joint tissues.²⁰ In brief, transect lines that are parallel to the middle line of the bones, and that are distributed along the entire bone area with an interval of 0.5 mm, were chosen to interpolate the optical absorption profile for each sectional slice. The interpolated absorption curve for each transect line was divided

FIG. 1. (a) Photograph of the transducer array/finger interface of the 3-D qPAT system and (b) the photographic section schematic of the finger joint imaging used in the current study. S: light source; T1–T8: transducer elements.

into three different joint tissues (bone, cartilage, and fluid) and the interface points among adjacent joint tissues were taken as 30% of the nearest maximum values along the curve. By using this FW30%M threshold, the mean thickness of the joint and the averaged optical absorption coefficients of joint tissues (bone, cartilage, and fluid) can be calculated in each sectional slice. The volumetric mean values of the joint thickness and the optical absorption coefficients of joint tissues can be further obtained by averaging the corresponding values in all sectional slices. The volumetric mean absorption coefficient of different joint tissues, the structural size, and the lowest sectional absorption coefficient of the joint cavity from both OA joints and healthy controls are used as indicators/markers for differentiation of OA from normal joints in this study. These indicators/markers have been previously employed by radiography or pure optical imaging methods to detect osteoarthritis/joint rheumatism.^{5,6,20,21}

II.B. Tissue phantom

Optically well-characterized tissue phantoms were used in our experiments to validate the accuracy of our 3D qPAT system before we conducted the *in vivo* experiments. The joint-like tissue mimicking phantoms were made of Intralipid, India ink, distilled water, and agar powder, among which Intralipid acts as the optical scatterer, India ink offers color contrast (optical absorber), and the agar (2%) serves as the coagulator. The procedure to prepare the optical tissue phantom has been previously described in detail,⁴⁵ and variations of optical absorption/scattering coefficients were very small in fabricated homogenous tissue phantom. To simplify the fabrication procedure in our study, two smaller cylindrical “bones” separated with an even joint space were embedded in a larger cylindrical background phantom to mimic a real finger joint. The joint space filled with “cartilage” was 1.5 mm thick, and had an absorption coefficient, μ_a of 0.01 mm⁻¹, and a reduced scattering coefficient, μ'_s of 1.5 mm⁻¹. The cylindrical background phantom had $\mu_a = 0.007$ mm⁻¹ and $\mu'_s = 1.0$ mm⁻¹, respectively, while the two fabricated “bones” had $\mu_a = 0.07$ mm⁻¹, $\mu'_s = 2.5$ mm⁻¹, and a diameter of 6 mm. The optical absorption/scattering coefficients of the joint-like phantoms well approximate the corresponding coefficients of *in vivo* finger joints.

II.C. Patients and clinical data

Seven subjects (two OA patients and five healthy controls) were enrolled in the current study between 2008 and 2009. All participants (white female; mean age 61 years, range 45–71 years) provided informed consent as part of the protocol approved by the Institutional Review Board of University of Florida (UF). Participants were recruited from the Rheumatology Clinic at the Shands Health Center of UF. One DIP finger joint, joint II of the left hand, from each subject was examined clinically and photoacoustically in this study. The DIP joint is mostly vulnerable to OA disease and easily accessible with our current 3-D photoacoustic scanning system.

Clinical examination of each patient was performed independently by a single rheumatologist. Patients with OA were identified by clinical history and main clinical features, including symptoms (predominantly pain and stiffness), functional impairment, and signs (joint enlargement and redness). The healthy controls had no known OA or other joint diseases.

The patients clinically examined were then photoacoustically scanned by our 3-D qPAT scanner and data processing were followed for image reconstruction. Data from six of the seven recruited subjects (2 OA and 4 healthy) resulted in successful photoacoustic image reconstruction, while data from one healthy control was not useable due to the severe motion error of finger during the photoacoustic examination.

III. RESULTS

Phantom results were examined before the *in vivo* joint experiments to validate the accuracy of our 3D qPAT system in a well controlled setting. The FW30%M method was used to analyze the reconstructed optical absorption profiles, where the interface between bone and cartilage (joint space) was determined by 30% of the maximum values on the optical absorption curve. Figures 2(a) and 2(b) show the reconstructed phantom joints in three-dimensional view, and in sagittal and coronal sections with our 3-D qPAT. The distributions of the recovered absorption coefficient along a transect crossing the bones and cartilage for the images shown in Figs. 2(a) and 2(b) are shown in Fig. 2(c). The lowest sectional absorption coefficient of the joint cavity when the cartilage had an absorption coefficient of 0.01 or 0.03 mm⁻¹ was found to be 0.012 and 0.026 mm⁻¹, respectively, where the reconstruction error is in the range of 13%–20% compared to the exact value. The volumetric mean absorption coefficient of the recovered joint cavity for the two phantom cases was found to be 0.016 and 0.03 mm⁻¹, respectively. While an absorption difference of the 0.01/0.03 mm⁻¹ for the cartilages is observed in the volumetric mean absorption coefficient of the recovered joint cavity, the error for the low optical absorption case is relatively larger. As discussed before,⁴¹ the relative larger error for the low optical absorption cases is a result of the simple iteration procedure, where a “U” shaped distribution curve of optical absorption coefficients is recovered as a “V” shaped distribution and low optical absorption seems more sensitive to the small error in the calculation of the optical fluence or photon density. The volumetric mean absorption coefficient of the bones was recovered as 0.067 and 0.070 mm⁻¹ for the two phantom cases, which agree well with the actual value of 0.07 mm⁻¹. The recovered cartilage thicknesses were found to be 1.5 and 1.8 mm, where the maximum error is 0.3mm relative to the actual size.

Consecutive coronal and sagittal slices as indicated in Fig. 1(b) from the recovered 3-D images of *in vivo* examined joints are presented below. The exact locations of these selected slices may be slightly different from subject to subject. The coronal and sagittal slices for a normal joint are shown in Figs. 3(a) and 3(b). As can be observed from

FIG. 2. Reconstructed optical absorption coefficient images in the coronal ($z = 0$ mm) and sagittal ($x = 0$ mm) sections as well as in 3-D view for a joint-like phantom where the absorption coefficient of "cartilage" was (a) 0.01 and (b) 0.03 mm⁻¹ (b). (c) shows the recovered absorption coefficient distributions along a transect crossing the "bones" and "cartilage" ($x = 0.5$ mm) for the images in coronal section.

Figs. 3(a) and 3(b), the bones are clearly delineated from the adjacent tissues (cartilage and fluid) in the joint cavity, which is clearly identified in both the coronal and sagittal sections. Further observation finds that the synovial fluid seems to be differentiable from the surrounding cartilages in some slices ($Z = 7.5$ mm, $Z = 8$ mm in the coronal section, and almost all the slices in the sagittal section). Figures 4(a) and 4(b) display selected coronal and sagittal slices from the recovered 3-D images for an OA joint. Again, the bones are clearly differentiable from the adjacent joint tissues (cartilage and fluid), and the joint cavity is easily identified in both the coronal and sagittal sections.

The average absorption coefficient and structural size of joint tissues for all the subjects are calculated and presented in Tables I. We note from Table I that the recovered volumetric mean absorption coefficient of the diseased synovial fluid ranges from 0.022 to 0.023 mm⁻¹, while it is below 0.017 mm⁻¹ (0.008–0.017 mm⁻¹) for the normal fluid. Difference between OA joints and healthy controls is also apparent, in terms of the volumetric mean absorption coefficient of the synovial fluid in the joint cavity. The difference is more striking when the ratio of the volumetric mean absorption coefficient of the synovial fluid to that of the bone is considered (0.58–0.63 for OA joints versus 0.25–0.46 for

TABLE I Averaged absorption coefficient and size of joint tissues for 2 OA and 4 normal joints.

	No.	Joint tissues	Mean μ_a (mm ⁻¹)	$\tilde{\mu}_a^a$ (mm ⁻¹)	Joint thickness (mm)	$\frac{\mu_a^{(c)}}{\mu_a^{(b)}}$	$\frac{\mu_a^{(f)}}{\mu_a^{(b)}}$	$\frac{\tilde{\mu}_a}{\mu_a^{(b)}}$
Healthy	1	Bone	0.035	0.004	1.7	0.43	0.25	0.11
		cartilage	0.015					
		fluid	0.008					
	2	Bone	0.039	0.010	2.1	0.61	0.45	0.26
		cartilage	0.024					
		fluid	0.017					
	3	Bone	0.034	0.009	2.5	0.59	0.45	0.26
		cartilage	0.020					
		fluid	0.015					
	4	Bone	0.037	0.008	1.8	0.61	0.46	0.22
		cartilage	0.023					
		fluid	0.017					
OA	5	Bone	0.036	0.018	1.5	0.76	0.63	0.50
		cartilage	0.028					
		fluid	0.023					
	6	Bone	0.037	0.014	1.1	0.72	0.58	0.38
		cartilage	0.026					
		fluid	0.021					

^a $\tilde{\mu}_a$ is the lowest sectional optical absorption coefficient in the joint cavity.

normal joints). For the diseased cartilage, the volumetric mean absorption coefficient (0.026–0.028 mm⁻¹) is generally larger than that of normal joints (0.015–0.024 mm⁻¹). The ratio of the volumetric mean absorption coefficient of the cartilage to that of the bone further confirms the difference between OA and normal joints (0.72–0.76 for OA joints versus 0.43–0.61 for normal joints).

Further observation from Table I indicates that the lowest sectional absorption coefficient in the joint cavity ranges from 0.014 to 0.018 mm⁻¹ in the diseased joints, while it is in the range of 0.004–0.010 mm⁻¹ for the normal joint. A more significant difference between the OA and normal joints is observed in the ratio of the lowest sectional absorption coefficient of the joint cavity to the volumetric mean absorption coefficient of the bone (0.38–0.50 for OA joints versus 0.11–0.26 for normal joints). The difference was found to be as large as 2:1 in the OA joints compared to the normal joints.

We also note from Table I that the measured thickness of the joint cavity (synovial fluid and cartilage) for OA joints ranges from 1.1 to 1.5 mm, with a mean value of 1.3 mm, whereas the range of this size for healthy joints is from 1.7 to 2.5 mm, with a mean value of 2.0 mm. Compared with the normal joints, the dimensional narrowing of the joint cavity is obvious for OA joints, which is in agreement with typical radiographic evidences of OA disease.^{5,6}

IV. DISCUSSION

In this study, for the first time we *in vivo* imaged distal interphalangeal joints and associated osteoarthritis in submillimeter resolution using our three-dimensional quantitative photoacoustic tomography approach. The joint cavity is clearly identified in both the coronal and sagittal

sections for examined DIP joints. Apparent differences, in both the reconstructed size and optical absorption coefficient of the joint cavity, are observed between osteoarthritic and normal joints. It is interesting to note that the spatial distribution of absorption coefficient appears to be heterogeneous for the OA joint, while it is relatively homogeneous for the healthy joint. For example, the shape and the optical absorption of the joint tissues (bone, cartilage, and synovial fluid) both change slowly as the sectional plane moves from one side of the examined DIP joint to another side for the healthy case (Fig. 3), while the corresponding spatial changes for the OA case (Fig. 4) are significantly larger. This observation is consistent with the findings from the previous optical imaging study.²⁰ As such, the optical absorption in some of the sectional slices is lower for the OA joint than that for the healthy joint, due to the heterogeneous absorption distribution for the OA joint.

Photoacoustic imaging of the joints offers a submillimeter spatial resolution and has the potential to be further enhanced when larger amount of transducer elements and higher frequency transducers are used. The joint space narrowing observed by photoacoustic examination is in agreement with typical radiographic evidences of OA disease, and the observed optical changes associated with diseased joints are consistent with documented biological roots. For example, it is known that with the onset of OA, the synovial membrane/fluid in articular cavity becomes increasingly turbid and has an enhanced vascularization.⁴⁶ The increased turbidity and vascularization would accompany increased optical absorption coefficient (as well as optical scattering coefficient) in the diseased synovial membrane/fluid. In fact, this optical increase can be as large as 100% at certain wavelengths in the NIR region.^{47,48} In addition, there is increasing evidence that OA is a disease involving a metabolic

FIG. 3. (a) Selected coronal and (b) sagittal planes of the recovered 3D absorption coefficient image for a typical normal joint. DP: distal phalanx; IP: intermediate phalanx; DIP: distal interphalangeal joint; CL: cartilage; SF: synovial fluid.

FIG. 4. (a) Selected coronal and (b) sagittal planes of the recovered 3D absorption coefficient image for a typical OA joint. DP: distal phalanx; IP: intermediate phalanx; DIP: distal interphalangeal joint; CL: cartilage; SF: synovial fluid; TE: tendon.

dysfunction of bone.^{49,50} It is likely that this metabolic dysfunction of bone, often associated with high metabolism of subchondral bone, will cause changes in its tissue optical absorption due to the changes in oxygenation, Hb, HbO₂, and water content.

An ultrasound array composed of eight 1 MHz transducers was used in the current study, which was sufficient to image the abnormalities in the joint cavity and the adjacent bones, where cartilage erosion has been generally believed to be the major cause of OA disease. It is worth to note that the ultrasound signals generated from the optical absorption of tissue have a broadband of frequency,⁵¹ which is different from the narrow band ultrasound signals in ultrasound imaging where the source ultrasound signals have almost the same frequency spectrum as the receiving ultrasound signals. Theoretical analysis has shown that the resolution of photoacoustic imaging is around $0.6 - 0.8\lambda_c$, where λ_c is the wavelength of photoacoustic signals with a high cutoff frequency of f_c .⁵¹ As such, photoacoustic imaging may provide better resolution over ultrasound imaging even when a same transducer is used. For example, the resolution can be around 0.45–0.6 mm for a high cutoff frequency f_c of 2 MHz, which is still detectable by the 1 MHz transducer with 80% bandwidth at –6 dB. As such, the imaging resolution of our current 3-D PAT approach and the uncertainty of measuring the joint thickness are around 0.5 mm, which can be further improved by the enhancement of the 3-D PAT technique. By using 1 MHz transducers, the joint cavity can be clearly identified and be measured in this study. As shown previously in photoacoustic imaging of cadaver joints,²² PAT was capable of imaging other interesting joint structures as well, including volar plate, tendon, and aponeurosis, when a high frequency transducer was used. In a recent study of whole-body mouse imaging using 3-D PAT,³⁶ both relatively large organs in centimeter size such as liver, kidney, and spleen, and small structures in millimeter size such as spinal cords, ovarian vessel, abdominal aorta, and femoral veins were clearly imaged when multiscale wavelet filter was used along with a 64-element 3.1 MHz ultrasound array. With these reported advancements in 3-D PAT technique, we believe that our 3-D PAT approach can be further improved to capture other interesting joint tissues such as ligaments, tendon, and synovium in addition to the articular cavity and the bones. This adds the potential for our 3-D PAT technique to detect OA that is initiated by these other joint tissues.^{52,53} Moreover, a detecting array with larger amount of transducer elements will also allow us to photoacoustically scan the finger joints with multispectral light in a short time, which extend the potential of 3D qPAT approach to quantify the neoangiogenesis and hypoxia in diseased finger joints.

It is also worth to note that PAT imaging is governed by ultrasound propagation behavior in tissue while ultrasound signals are excited by optical pulses. For simplicity, in this study, we have assumed an acoustically homogenous medium in image reconstruction. The actual inhomogeneous nature of joint tissues will definitely affect the ultrasound propagation behavior in computation. In this case, an advanced reconstruction algorithm considering acoustical

heterogeneity is needed for improved image reconstruction. Such an advanced algorithm can also reconstruct tissue acoustic properties such as ultrasound speed in addition to tissue optical properties.^{43,54} While the advanced reconstruction algorithm requires increased computation cost, and the recovered tissue acoustic properties together with tissue optical properties and anatomical structure of joints tissues will offer more accurate detection of OA disease. For example, decrease of ultrasound speed and increase of ultrasound attenuation have been observed in osteoarthritic cartilage.^{55,56}

V. CONCLUSIONS

In summary, this study represents the first attempt to *in vivo* detect osteoarthritis in the finger joints using three-dimensional quantitative photoacoustic tomography. As an emerging imaging technique, the advantage of 3-D qPAT for early detection of OA lies in its capability to quantitatively image optical absorption coefficient, acoustical properties, and physiological/functional parameters such as oxygenation and water content as well as to reveal joint abnormality in high resolution. In addition, PAT is portable and low in cost, and hence may play an important role in long term monitoring of OA. With this study, we show that apparent differences, in both the reconstructed size of the joint space and the absorption coefficient of the joint cavity, exist between OA and normal joints examined by three-dimensional photoacoustic imaging technique. The results reported here suggest the possibility of 3-D qPAT as a potential clinical tool for early detection OA in the finger joints. We are encouraged to further evaluate the 3-D qPAT approach and to improve our imaging system for extended capabilities in the near future. For example, we will compare the diagnosis performance of our 3-D qPAT approach with routinely used plain radiography or MRI by performing a statistical study on a sample of osteoarthritic joints. We also plan to adopt a larger transducer array with higher frequency, which we believe will allow us to image more fine structures in joint tissues and to quantify the neoangiogenesis and hypoxia in the diseased finger joints.

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