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Kinetic $\text{[^{18}F]}$-Fluoride of the Knee in Normal Volunteers

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**Purpose:** $\text{[^{18}F]}$-sodium fluoride ($\text{[^{18}F]}$NaF) is a well-established bone-seeking agent that has shown promise to assess bone turnover in a variety of disorders, but its distribution in healthy knee joints has not been explored. This study aimed to investigate parametric values for $\text{[^{18}F]}$NaF uptake in various bone tissue types of the knee and their spatial distributions.

**Methods:** Twelve healthy subjects were hand-injected with 92.5 MBq of $\text{[^{18}F]}$NaF and scanned on a 3-T PET/MRI system. Listmode PET data for both knees were acquired for 50 minutes from injection simultaneously with MRI Dixon and angiography data. The image-derived input function was determined from the popliteal artery. Using the Hawkins model, Patlak analysis, data interpretation, writing, or decision to submit the manuscript. F.K., N.R.J. (0000-0001-9624-5210), C.S. (0000-0001-8063-6508), and F.K. (0000-0001-7529-1763) have no conflicts of interest to declare.

**Conclusions:** Quantitative $\text{[^{18}F]}$NaF PET is sensitive to variations in bone vascularization and metabolism in the knee joint.

**Key Words:** bone, fluoride, hybrid imaging, kinetics, knee, MRI, NaF, PET, PET/MRI

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The coils were used because of their lower attenuation effect on PET photons. Both knees were scanned using 1 flexible phased-array receive-only coil (NeoCoil, Pewaukee, Wis) around each knee. The coils were used because of their lower attenuation effect on PET photons. Both knees were scanned using 1 flexible phased-array receive-only coil (NeoCoil, Pewaukee, Wis) around each knee.

PET/MRI Scanning

Subjects were scanned on a 3-T whole-body time-of-flight PET-MR hybrid system (GE Healthcare, Milwaukee, Wis). Each subject was positioned feet first with a 16-channel flexible phased-array receive-only coil (NeoCoil, Pewaukee, Wis) around each knee. The coils were used because of their lower attenuation effect on PET photons. Both knees were scanned using 1 flexible phased-array receive-only coil (NeoCoil, Pewaukee, Wis) around each knee.

Magnetic resonance angiography data were acquired using a 3-dimensional gradient echo sequence with imaging parameters: repetition time/echo time (TE) = 21/2.1 milliseconds, slices = 18, slice thickness = 1.2 mm, and flip angle = 15°. A 2-point Dixon fat-water T1-weighted spoiled gradient echo MR sequence was acquired for MR-based attenuation correction of PET data with acquisition parameters: repetition time/TE1/TE2 = 4.1/1.1/2.2 milliseconds, field of view = 50 × 37.5 cm, matrix = 256 × 128, slice thickness/overlap = 5.2/2.6 mm, 120 images/slab, scan time = 18 seconds.

For calculation of the image-derived input function (IDIF), dynamic PET frame times of 40 × 1 seconds, 13 × 10 seconds, and 23 × 2 minutes were reconstructed using time of flight–ordered subset expectation maximization with 3 iterations and 21 subsets with corrections for decay, attenuation, scatter, random, and dead time. Time activity curves of bone PET uptake were determined using dynamic PET time frames of 8 × 2 seconds and 24 × 2 minutes with the same corrections. A 3-mL venous blood sample was taken 50 minutes after injection when arterial and venous blood concentrations have equilibrated and measured in a well counter.

IDIF Calculation

The IDIF was determined from $^{18}$F-NaF activity (kBq/mL) within the popliteal artery of each knee. The artery was segmented automatically from MR angiography images and a short time-frame PET angiogram during the arterial phase (0–16 seconds after injection) when the tracer is predominantly in the arteries. In order to minimize spillover artifacts, the voxels centered in the middle of the artery were determined for each dynamic PET frame and used for the IDIF. Central voxels were defined by including the voxels in each axial slice within the highest 10% of arterial NaF activity.

Bone Segmentations and Regions of Interest

Using the in-phase, out-of-phase, and water images from the Dixon scans, masks covering the femur, patella, and tibia were first created by manually drawing regions of interest (ROIs) (Fig. 2). The bone tissue was then segmented further to create subchondral/cortical bone masks using k-means clustering (4 cluster groups minimized to squared Euclidean distance repeated 4 times with different initial centroids). The long bone of the femur and tibia was identified as the cortical bone 6 to 8 cm from the center of the joint space. Trabecular bone ROIs in the tibia and distal end of the femoral bone were drawn for both legs maintaining a minimum of 3-mm distance from the edge of the bone to avoid partial voluming. Thereafter, the subchondral bone of the femur was manually subdivided into 5 regions: trochlea and central and posterior regions of the medial and lateral compartments. Similarly, tibial subchondral bone was further separated into lateral and medial regions. Lastly, cortical bone at the site of a patellar tendon insertion (tibial tuberosity) was identified and excluded from the analysis of cortical bone.

SUV and Kinetic Modeling

The time activity curves and IDIF data were fitted to the 2-tissue compartment model using the Patlak method and again using the non-linear regression (NLR) method. Patlak analysis is a graphical technique for estimating $K_p$, the total rate of plasma clearance of NaF to the bone matrix, which assumes that $^{18}$F is irreversibly bound to bone mineral ($k_4 = 0$). Data from 10 to 50 minutes were fit to allow for equilibration between tracer in plasma and the bone extracellular...
fluid. Nonlinear regression fitting included estimated 3 rate parameters ($K_1$, $k_2$, and $k_3$) along with a partial volume fraction, a blood fraction, and an input dispersion estimate and was computed using COMKAT software. The rate constant $k_4$ was predefined as 0. For both blood fraction and $K_1$, a parameter range from 0 to 1 was applied, whereas a range of 0.015 to 0.8 was used for $k_2$ and $k_3$, and 0 to 2 seconds for the dispersion constant tau. To avoid local minima, fits were repeated with 3 starting conditions, and results with the lowest residuals were used. The rate of total plasma clearance using the NLR method ($K_{\text{NLR}}$) was calculated from the $K_1$, $k_2$, and $k_3$ values obtained by using the following formula:

$$K_{\text{NLR}} = K_1 (k_3 / (k_2 + k_3))$$  (1)

$K_{\text{NLR}}$, like $K_1$, has units of mL·min$^{-1}$·mL$^{-1}$, whereas $k_2$ and $k_3$ have units of min$^{-1}$.

The $K_{\text{NLR}}$ parameter can be separated into 2 parameters of physiological interest. One parameter is $K_1$, the rate of transit of the $^{18}$F plasma concentration into the extravascular compartment, and reflects flow delivery of the tracer. Perfusion ($P$) estimates for each ROI were derived from $K_1$ values using least squares regression to the Renkin-Crone formula:

$$K_1 = F^{-1} (1 - \exp(-PS/F))$$  (2)

where the product of permeability and surface area (PS) was assumed to be 0.24, as reported by Pieri et al. The second physiological parameter is the extraction fraction, $k_3/(k_2 + k_3)$, which represents the fraction of $^{18}$F entering the tissue that binds to the bone matrix as opposed to reentering the bloodstream.

Images for mean SUV and SUV max were calculated from images obtained by averaging the last 2 frames of the dynamic study (46–50 minutes).

Statistical Analysis

Values across the entire patient cohort are reported as median with interquartile range, and P values are from paired Student 2-tailed t-tests using a threshold for significance of $P < 0.05$ after Bonferroni correction for multiple comparisons. Correlations between obtained parameters were analyzed using least products linear regression where goodness of fit was evaluated with a Pearson adjusted $R^2$ value. Reproducibility between IDIF blood activity and venous blood samples was analyzed by calculating the coefficient of variation, reported in percent. Image coregistration, ROI analysis, calculations, and statistical analysis were performed with software created in MATLAB 2013b (MathWorks, Natick, Mass).

RESULTS

Median parametric values along with interquartile range across all subjects are presented in Table 1. Variations in global $[^{18}$F$]$NaF uptake were observed between subjects (Fig. 3) with consequently higher or lower SUV values across all 3 types of bone tissue. Comparisons between the 3 bone tissue types are shown in Figure 4. Cortical bone had highest $[^{18}$F$]$NaF uptake for all measured parameters compared with trabecular bone ($P < 0.01$), which had the lowest uptake. SUV and $K_1$ values for subchondral bone were lower than that of cortical bone, but these differences were not significant after correction for multiple comparisons. Subjects had significantly higher SUV max and $K_1$ values and a significantly lower extraction fraction in cortical bone compared with subchondral bone. Subchondral bone had significantly higher $[^{18}$F$]$NaF uptake (SUV, SUV max, and $K_{\text{NLR}}$; $P < 0.01$) than trabecular bone tissue.

There was a regional variance in distribution of $K_1$ and extraction fraction values. The distribution ranged from cortical bone...
of the shaft, which had the highest vascularization where $K_1 > K_i$ and $k_3/(k_3 + k_2) < 1$, to the trochlea and patella region of subchondral bone, where $k_3/(k_3 + k_2) \approx 1$ and $K_1 \approx K_i$ (Fig. 5). By visual analysis of $K_1$ and $k_3/(k_3 + k_2)$ maps, a negative gradient of $K_1$ values can be seen from the femoral and tibial shafts decreasing toward the joint space. A second gradient can be seen as $K_1$ is higher in subchondral bone and declines toward the center of the trabecular bone of the femur and tibial head (Fig. 6). In the subchondral bone of the femur, $K_1$ and blood volume values were higher in the posterior section, decreasing to the lowest in the trochlea ($P < 0.01$). The

FIGURE 3. SUV of representative slice from 2 subjects. There was a wide intersubject range of [$^{18}$F]NaF uptake across the joint. The left image is a subject with low uptake in knee, whereas the right image is a subject with high uptake in all bone tissues. In addition to global variations in tracer uptake between subjects, some individual variations in the relative distribution of [$^{18}$F]NaF uptake across bone tissues regions were observed. In this example, the subject on the right has relatively low uptake in the subchondral bone of the femur compared with the subchondral bone of the patella and tibial head, whereas the subject on the left has equally low uptake in all subchondral regions.

FIGURE 4. Parametric values of [$^{18}$F]NaF uptake for different bone tissue types of the knee. Cortical bone had the highest [$^{18}$F]NaF uptake in all measured parameters when compared with trabecular bone, which had the lowest uptake. Subchondral bone also had higher uptake than trabecular bone with significantly higher SUVmean, SUVmax, and $K_{\text{NLR}}$ values, yet only slightly elevated $K_1$ and $k_3/(k_3 + k_2)$ values. The relative distribution of $K_{\text{NLR}}$ values between bone tissues was almost identical to that of SUV. Note that despite having the highest uptake as expressed by SUV and $K_{\text{NLR}}$, cortical bone has the lowest extraction fraction. P values were corrected for 18 comparisons using a Bonferroni correction ($†P < 0.01$ difference compared with cortical bone, $‡P < 0.01$ difference between trabecular bone and subchondral bone).
opposite gradient was observed in the extraction fraction maps, resulting in total metabolism $K_{NLR}^3$ and SUV images that were more spatially uniform.

There were no other significant differences in $[^{18}F]$NaF uptake parameters between the 3 bone tissue types or between subregions of subchondral bone. The sites of tendon insertion had elevated SUV values and significantly higher $K_{NLR}^3$ ($P < 0.05$) than remaining cortical bone tissue. The vascularization was lower ($K_1$ 33% less, blood volume 84% lower [both $P > 0.01$]), and the extraction fraction higher ($k_3/(k_3 + k_2)$ 88% higher, $P < 0.01$).

$K_{NLR}^3$ values correlated highly with SUV values ($R^2 = 0.90$). $K_{NLR}^3$ values from the Patlak method had a slightly poorer correlation to SUV ($R^2 = 0.87$) and were 17% lower than those obtained by NLR (Fig. 7). The correlation of $K_{NLR}^3$ values to $K_{NLR}^3$ values was high ($R^2 = 0.97$) despite the 17% bias (Fig. 7). Using $K_1$ values from the NLR fit, flow values were obtained and found to be within a few percentage points of $K_1$ values (Table 1). The $K_1$ values were in the range where $K_1 < < PS$, and thereby the condition $F \approx K_1$ applies.

Group average IDIF values at 1, 5, 10, and 50 minutes were 10.2, 6.0, 4.2, and 2 kBq/mL when normalized to a 100-MBq injection. At 50 minutes, mean IDIF values were 6% higher than mean venous blood sample values. Coefficient of variation values between venous blood samples and IDIF values measured at 50 minutes were 8.3%. Repeated injections in 1 subject had mean coefficient of variation values of 9% across time points observed between 1 and 50 minutes (Fig. 8).

**DISCUSSION**

Semiquantitative and quantitative values for $[^{18}F]$NaF uptake in the knee were obtained from healthy subjects using PET/MRI. A large intersubject variation in NaF uptake was observed as there were significant differences in uptake parameters between cortical bone and the subchondral/trabecular bone tissues. Trabecular bone was found to have significantly lower SUV, $K_{NLR}^1$, and blood volume values yet a significantly higher extraction fraction than the cortical bone tissue in the shaft of the femur and tibia. Blood volume was the parameter with the largest discrepancy between bone tissues being significantly higher in the shaft compared with subchondral or trabecular bone of the knee. Subjects had higher vascularization (larger blood volume and higher $K_1$ values) in the shaft of the femur and tibia declining with a negative gradient toward the joint space reaching the lowest values at the center of the trabecular bone near the distal end. This $K_1$ gradient was partially offset by a gradient of increasing extraction efficiency that was significantly lower in the shaft. A similar regional discrepancy was also evident in SUV, SUVmax, and $K_1$ values, although to a lesser degree. These parameters, like $K_1$, were significantly higher in the shaft decreasing in the subchondral bone and trabecular bone of the knee joint in these healthy individuals. Likewise, the sites of tendon insertion of the cortical bone had much lower vascularization ($K_1$ and blood volume), yet a net uptake than regular cortical bone due to a high extraction fraction. A similar observation has been
made between the spine and humeral bone tissues where low $K_1$ values in the humeral bone were partially compensated by a higher $k_3/(k_2 + k_3)$ to give a more comparable, yet still significantly lower, $K_1$ value.\textsuperscript{22,23} Aside from the $K_1$ and $k_3/(k_2 + k_3)$ gradients, all other parametric values within the subchondral bone tissue ROIs of subjects were quite homogenous with no significant differences when comparing subchondral subregions across the patella, femur, and tibia. $K_i^{\text{Patlak}}$ values from the Patlak method were 17% lower than those obtained by NLR, which is a larger bias than previously reported by Siddique et al.,\textsuperscript{17} where $K_i^{\text{Patlak}}$ was 10% lower than $K_i^{\text{NLR}}$ in the lumbar vertebrae. Still, $K_i^{\text{NLR}}$ values correlated highly with both $K_i^{\text{Patlak}}$ values ($R^2 = 0.97$) and SUV ($R^2 = 0.90$) with no regional variations in their correlation. Ultimately, this study gives no evidence of meaningful differences in using Patlak or NLR methods to determine $K_i$ as they could be interchanged with a conversion factor. Studies including mean SUV, $K_i^{\text{Patlak}}$, and $K_i^{\text{NLR}}$ have found these parameters to have similar reproducibility with coefficients of variation ranging between 9% and 15%\textsuperscript{17,21,23,25} although $K_i^{\text{NLR}}$ had lower reproducibility when $k_1$ is not limited to 0 when fitting. In this study, SUV, $K_i^{\text{Patlak}}$, and $K_i^{\text{NLR}}$ have comparable variance where intersubject SDs are between 43% and 46% of mean values. Despite similar reproducibility, $K_i$ values have been reported to be a more sensitive measure of regional bone metabolism than SUV.\textsuperscript{17,18,27,42} In the limbs, where F\textsuperscript{+} uptake is low, Brenner et al.\textsuperscript{48} and Apostolova and Brenner\textsuperscript{44} and concluded the minimal change of SUV in a patient must be greater than 50% to reliably detect disease or treatment-related changes, whereas the same diagnosis could be made from a change in $K_i$ of 25%. $K_i$ values have also shown to be more sensitive when analyzing alterations in subchondral bone of the femur adjacent to cartilage defects.\textsuperscript{4} SUVmax values in this study are similar to previously reported mean SUVmax values of 2.44 for the tibia\textsuperscript{43} and 2.22 in the femur shaft.\textsuperscript{44} SUVmax has been found to correlate well with adjacent cartilage alterations,\textsuperscript{24} and although it had the largest intrasubject variation in this study, it had a relatively lower variance between subjects and greater differentiation between bone tissues (Fig. 4). In this study, using NLR was advantageous as obtaining $K_1$ and extraction fraction parameters provided useful information that could not be extracted from $K_i$ alone.

The $K_1$ values obtained in this study were within a flow-dominant regimen where it has, theoretically, a linear correlation to blood flow ($K_1 < < \text{PS}$). The flow values obtained in this study compare well with measured blood flow in the femoral shaft,\textsuperscript{45} but lack a criterion-standard measure to investigate $K_1$ as a surrogate flow measure. To date, the most convincing studies to confirm the relationship between blood flow in bone tissue and $K_1$ for [\textsuperscript{18}F]NaF kinetics have been performed in swine vertebrae.\textsuperscript{15,21} Since then, authors have reported a poor correlation between $K_1$ and bone perfusion in studies of the hip of human surgery patients\textsuperscript{46} and the forelimbs of healthy rats.\textsuperscript{46} Obtaining an estimate of flow would be of great clinical value. Bone perfusion is usually linked to metabolic activity and varies greatly between different bones and bone regions in the skeleton where the extremities are among the lowest.\textsuperscript{23,25,47}

FIGURE 6. Example of the distribution of $K_1$, $K_1^{\text{NLR}}$ and SUV from 1 subject. Subjects had a negative gradient of $K_1$ values from the shaft toward the joint space and from the subchondral bone toward the center of the distal end of the femoral bone and tibial head. In the sagittal plane, $K_1$ values were highest in the posterior section of subchondral bone and decreased in the anterior direction toward the trochlea and patella. However, the opposite gradient was observed in extraction fraction $k_3/(k_2 + k_3)$ maps, resulting in $K_1^{\text{NLR}}$ and SUV images with more localized heterogeneity.
Perfusion studies using microspheres have shown a reduction of blood flow in bones related to age, osteoporosis, and reduced endothelium-dependent vasodilation.

With regard to this study’s aim to report key parametric values for $^{18}F$NaF uptake in the healthy knee, there are several limitations to be considered when interpreting the results. First, the number of subjects is small where results can be skewed by relatively few abnormalities. The range of ages (22–44 years) is a period of rather stable bone density in human adults, but factors such as body mass index, varus/valgus alignment, disease, or activity level could alter the kinetics in bone tissue. Second, despite the numerous advantages from combining PET imaging with MRI in knee examinations, there are disadvantages in foregoing the superior information on bone density, which CT provides. This information is valuable in both the attenuation correction of PET data and the segmentation of bone tissue. Dixon-based methods, as used by the scanner in this study, have been shown to underestimate bone $^{18}F$NaF mean SUV by 10%, ranging between 0% and 20% depending on location. The subchondral bone would be least affected being close to the bone surface, whereas the trabecular bone could have a more pronounced underestimation of SUV due to improper attenuation correction. Likewise, a similar underestimation of $K_i$ and $K_t$ would be expected, although it would be partially offset by a similar underestimation of activity in input function obtained from the popliteal artery. Lastly, the use of an IDIF would best be confirmed by using arterial sampling as a criterion standard. In this study, venous samples confirmed the activity of the later phase of the IDIF but not the earlier phase of high activity.

PET/MRI is an optimal dual-imaging combination offering the advantages of the high soft tissue contrast and resolution of MRI and the sensitivity of PET. In this study, MR angiography added the advantage of segmenting the popliteal artery, making an automated process to obtain the IDIF possible. The input functions obtained correspond well with literature values for $^{18}F$NaF from arterial sampling, and visual inspection of generated ROIs confirmed successful automated segmentation of the popliteal artery. Mean IDIF values 50 minutes after injection were 6% higher than venous blood samples taken on an equilibrium time point, whereas Cook et al. found arterial blood samples to be 2% higher than venous blood samples after 24 minutes. With the increased use of NaF in nononcological studies of the skeleton, it has become even more relevant as moderate differences in NaF uptake may be an early indication of bone degradation in diseases such as osteoarthritis. The combination of PET/MRI reduces the radiation dose significantly in 2 ways. First by eliminating CT and, second, because the PET data are acquired for the duration of the MRI protocol (which can be up

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**FIGURE 7.** Comparison of $K_{NLR}^\text{ROI}$ with SUV values and $K_{Pat}^\text{Kpat}$ Patlak. A, Scatterplot of $K_{NLR}^\text{ROI}$ results from all ROIs of all subjects plotted against ROI mean SUV ($R^2 = 0.90$). The Bland-Altman plot compares SUV values and $K_{NLR}^\text{ROI}$ values multiplied by the slope determined from the regression (slope = 0.90). B, A scatterplot with regression fit and Bland-Altman plot of the same ROIs comparing $K_{NLR}^\text{ROI}$ vs $K_{Pat}^\text{Kpat}$ values ($R^2 = 0.97$). The Patlak method produced $K_i$ values that were 17% lower than those obtained by NLR and had a slightly poorer correlation to SUV ($R^2 = 0.87$).
The effective dose of this study is estimated to be 2.16 mSv. and still retain the same signal-to-noise ratio in PET SUV maps. A standard clinical dose of 200 MBq to 90 MBq (used in this study) to an hour), the injected dose of $^{18}$F-fluoride can be decreased from a standard clinical dose of 200 MBq to 90 MBq (used in this study) and still retain the same signal-to-noise ratio in PET SUV maps. The effective dose of this study is estimated to be 2.16 mSv.26 Quantitative MRI techniques have been widely studied to develop robust biomarkers for the early detection and monitoring of osteoarthritis and monitoring of patients having had an anterior cruciate ligament injury.32

CONCLUSIONS

This study showed significant variations in regional bone perfusion and metabolism between skeletal tissue types in the knee joint. We have shown the feasibility of using PET/MR to create an accurate IDIF from the popliteal artery and to conduct a quantitative and semiquantitative evaluation of bone metabolism in the knee at a low radiation dosage. $^{18}$F-NaF PET/MRI is a noninvasive technique that offers an attractive tool to simultaneously estimate bone perfusion and metabolism at clinically relevant sites of the knee.

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REFERENCES

1. Jadvar H, Desai B, Conti PS. Sodium $^{18}$F-fluoride PET/CT of bone, joint, and other disorders. Semin Nucl Med. 2015;45:58–65.
2. Kogan F, Fan AP, Monu U, et al. Quantitative imaging of bone-cartilage interactions in ACL-injured patients with PET-MRI. Osteoarthr Cartil. 2018;26:790–796.
3. Kogan F, Broksi SM, Yoon D, et al. Applications of PET-MRI in musculoskeletal disease. J Magn Reson Imaging. 2018;48:27–47.
4. Savic D, Pediya V, See Y, et al. Imaging bone-cartilage interactions in osteoarthrits using $^{18}$F-NaF PET-MRI. Mol Imaging. 2016;15:1–12.
5. Kogan F, Fan AP, McWalter EJ, et al. PET/MRI of metabolic activity in osteoarthritis: a feasibility study. J Magn Reson Imaging. 2017;45:1736–1745.
6. Watanabe T, Takase-Minegishi K, Itaha A, et al. (18)F-FDG and (18)F-NaF PET/CT demonstrate coupling of inflammation and accelerated bone turnover in rheumatoid arthritis. Med Rheumatol. 2016;26:180–187.
7. Raynor W, Houshmand S, Gholami S, et al. Evolving role of molecular imaging with (18)F-sodium fluoride PET as a biomarker for calcium metabolism. Curr Osteoporos Rep. 2016;14:115–125.
8. Kobayashi N, Inaba Y, Tateishi U, et al. Comparison of $^{18}$F-fluoride positron emission tomography and magnetic resonance imaging in evaluating early-stage osteoarthrits of the hip. Nucl Med Commun. 2015;36:84–89.
9. Draper CE, Quon A, Fredericson M, et al. Comparison of MRI and $^{18}$F-NaF PET/CT in patients with patellofemoral pain. J Magn Reson Imaging. 2012;36:928–932.
10. Blau M, Ganatra R, Bender MA. $^{18}$F-fluoride for bone imaging, Semin Nucl Med. 1972;2:31–37.
11. Narita N, Kato K, Nakagaki H, et al. Distribution of fluoride concentration in the rat’s bone. Calcif Tissue Int. 1990;46:200–204.
12. Kato K, Nakagaki H, Robinson C, et al. Distribution of fluoride across cementum, dentine and alveolar bone in rats. Caries Res. 1990;24:117–120.
13. Ishiguro K, Nakagaki H, Taubli S, et al. Distribution of fluoride in cortical bone of human rib. Calcif Tissue Int. 1993;52:278–282.
14. Messa C, Goodman WG, Hoh CK, et al. Bone metabolic activity measured with positron emission tomography and $^{18}$F-fluoride ion in renal osteodystrophy: correlation with bone histomorphometry. J Clin Endocrinol Metab. 1993;77:949–955.
15. Pietr M, Zittel TT, Becker GA, et al. Assessment of porcine bone metabolism by dynamic J Nucl Med. 2001;42:1091–1100.
16. Hawkins RA, Choi Y, Huang SC, et al. Evaluation of the skeletal kinetics of fluorine-18-fluoride ion with PET. J Nucl Med. 1992;33:633–642.
17. Siddique M, Frost ML, Blake GM, et al. The precision and sensitivity of (18)F-fluoride PET for measuring regional bone metabolism: a comparison of quantification methods. J Nucl Med. 2011;52:1748–1755.
18. Brenner V, Verpon C, Muiz M, et al. Comparison of different quantitative approaches to $^{18}$F-fluoride PET scans. J Nucl Med. 2004;45:1493–1500.
19. Al-Bayatti Y, Siddique M, Frost ML, et al. Precision of $^{18}$F-fluoride PET skeletal kinetic studies in the assessment of bone metabolism. Osteoporos Int. 2012;23:2533–2541.
20. Frost ML, Blake GM, Park-Holohan SJ, et al. Long-term precision of $^{18}$F-fluoride PET skeletal kinetic studies in the assessment of bone metabolism. J Nucl Med. 2008;49:700–707.
21. Pietr M, Machulla H-J, Jahn M, et al. Coupling of porcine bone blood flow and metabolism in high-turnover bone disease measured by [(15)O][H][2]O and [(18)F]fluoride ion positron emission tomography. Eur J Nucl Mol Imaging Mol. 2002;29:907–914.
22. Rajmakers P, Temmerman OP, Saridin CP, et al. Quantification of $^{18}$F-fluoride kinetics: evaluation of simplified methods. J Nucl Med. 2014;55:1122–1127.
23. Frost ML, Blake GM, Cook GI, et al. Differences in regional bone perfusion and turnover between lumbar spine and distal humerus: (18)F-fluoride PET study of treatment-naive and treated postmenopausal women. Bone. 2009;45:942–948.
24. Nalumia C, Cockshott WP, Belbeck LW, et al. Measurement of absolute bone blood flow by positron emission tomography. Skeletal Radiol. 1986;15:198–200.
25. Cook GI, Lodge MA, Blake GM, et al. Differences in skeletal kinetics between vertebral and humeral bone measured by $^{18}$F-fluoride positron emission tomography in postmenopausal women. J Bone Miner Res. 2000;15:763–769.
26. Beheshi M, Mottaghy FM, Paycha F, et al. (18)F-NaF PET/CT: EANM procedure guidelines for bone imaging. Eur J Nucl Mol Imaging Mol. 2015;42:1767–1777.
27. Blake GM, Siddique M, Frost ML, et al. Imaging of site specific bone turnover in osteoporosis using positron emission tomography. Curr Osteoporos Rep. 2014;12:475–485.
28. Khalighi MM, Deller TW, Fan AP, et al. Image-derived input function estimation on a TOF-enabled PET/MR for cerebral blood flow mapping. J Cereb Blood Flow Metab. 2018;38:126–135.
29. Akella SV, Regatte RR, Gougoutas AJ, et al. Proteoglycan-induced changes in Ti-Ih0-relaxation of articular cartilage at 4T. Magn Reson Med. 2001;46:419–423.
30. Wheaton AJ, Casey FL, Gougoutas AJ, et al. Correlation of Ti-Ih0 with fixed charge density in cartilage. J Magn Reson Imaging. 2004;20:519–525.
31. Mosher TJ, Zhang Z, Reddy R, et al. Knee articular cartilage damage in osteoarthritis: analysis of MR image biomarker reproducibility in ACRIN-PA 4001 multicenter trial. *Radiology*. 2011;258:832–842.

32. Monu UD, Jordan CD, Samuelson BL, et al. Cluster analysis of quantitative MRI T2 and T1ρ relaxation times of cartilage identifies differences between healthy and ACL-injured individuals at 3T. *Osteoarthritis Cartilage*. 2016;25:513–520.

33. David-Vaudey E, Ghosh S, Ries M, et al. T2 relaxation time measurements in osteoarthritis. *Magn Reson Imaging*. 2004;22:673–682.

34. Haddock B, Holm S, Poulsen JM, et al. Assessment of muscle function using hybrid PET/MRI: comparison of 18F-FDG PET and T2-weighted MRI for quantifying muscle activation in human subjects. *Eur J Nucl Med Mol Imaging*. 2017;44:704–711.

35. Kogan F, Levine E, Chaudhari AS, et al. Simultaneous bilateral-knee MR imaging. *Magn Reson Med*. 2018;80:529–537.

36. Wagenknecht G, Kaiser H-J, Mottaghy FM, et al. MRI for attenuation correction in PET: methods and challenges. *MAGMA*. 2013;26:99–113.

37. Blake GM, Siddique M, Puri T, et al. A semipopulation input function for quantifying static and dynamic 18F-fluoride PET scans. *Nucl Med Commun*. 2012;33:881–888.

38. Lådeman L, Sreenivasa G, Michel R, et al. Corrections of arterial input function for dynamic H215O PET to assess perfusion of pelvic tumours: arterial blood sampling versus image extraction. *Phys Med Biol*. 2006;51:2883–2900.

39. Patlak CS, Blasberg RG, Fenstermacher JD. Graphical evaluation of blood-to-brain transfer constants from multiple-time uptake data. *J Cereb Blood Flow Metab*. 1983;3:1–7.

40. Mozic RF Jr, Cornelius S. COMKAT: compartment model kinetic analysis tool. *J Nucl Med*. 2001;42:636–645.

41. Renkin EM. Transport of potassium-42 from blood to tissue in isolated mammalian skeletal muscles. *Am J Physiol*. 1959;197:1205–1210.

42. Apostolova L, Bremer W. Measuring bone metabolism with fluoride PET: methodological considerations. *PET Clin*. 2010;5:247–257.

43. Sabbah N, Jackson T, Mosci C, et al. 18F-sodium fluoride PET/CT in oncology: an atlas of SUVs. *Clin Nucl Med*. 2015;40:e228–e231.

44. Win AZ, Aparici CM. Normal SUV values measured from NaF18-PET/CT bone scan studies. *PLoS ONE*. 2014;9:e108429.

45. Temmerman OP, Raijmakers PG, Kloet R, et al. In vivo measurements of blood flow and bone metabolism in osteoarthritis. *Rheumatol Int*. 2013;33:959–963.

46. Tomlinson RE, Silva MJ, Shoghi KI. Quantification of skeletal blood flow and fluoride metabolism in rats using PET in a pre-clinical stress fracture model. *Mol Imaging Biol*. 2012;14:348–354.

47. Bloomfield SA, Hogan HA, Delp MD. Decreases in bone blood flow and bone material properties in aging Fischer-344 rats. *Clin Orthop Relat Res*. 2002;248–257.

48. Griffith JF, Yeung DK, Tsang PH, et al. Compromised bone marrow perfusion in osteoporosis. *J Bone Miner Res*. 2008;23:1068–1075.

49. Prisby RD, Ramsey MW, Behnke BJ, et al. Aging reduces skeletal blood flow, endothelium-dependent vasodilation, and NO bioavailability in rats. *J Bone Miner Res*. 2007;22:1280–1288.

50. Aznar MC, Sersar R, Saabye J, et al. Whole-body PET/MRI: the effect of bone attenuation during MR-based attenuation correction in oncology imaging. *Eur J Radiol*. 2014;83:1177–1183.

51. Kogan F, Fan AP, Gold GE. Potential of PET-MRI for imaging of non-oncologic musculoskeletal disease. *Quant Imaging Med Surg*. 2016;6:756–771.