Fabrication of an anthropomorphic heterogeneous mouse phantom for multimodality medical imaging

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Abstract
This work presents a comprehensive methodology for constructing a tissue equivalent mouse phantom using image modeling and 3D printing technology. The phantom can be used in multimodality imaging and irradiation experiments, quality control, and management. Computed tomography (CT) images of a mouse were acquired and imported into 3D modeling software. A skeleton and skin shell models were segmented in the modeling software and manufactured using 3D printing technology. The bone model was constructed with VERO-WHITE printing material with additional ingredients, including a photosensitive resin, polyurethane epoxy resin, and acrylate. Acrylonitrile butadiene styrene resin material was used to construct the skin shell. The skin shell was attached to the skeleton and filled with a specially formulated gel to act as a soft tissue substitute. The gel consisted of agarose, micro-pearl powder, sodium chloride, and magnevist solution (gadopentetate dimeglumine). A micro-container filled with 18F-fluorodeoxyglucose (18F-FDG) radioactive tracer was placed in the abdomen for micro and human positron emission tomography (PET)/CT imaging. The mouse phantom had tissue equivalency in dose attenuation with x-rays and relaxation times with magnetic resonance imaging (MRI). The CT Hounsfield Unit (HU) range for the gel soft tissue material was 31–36 HU. The 3D printed bone mimetic material had equivalent tissue/bone contrast compared with in vivo mouse measurements with a mean value of 130 ± 10 HU. At different magnetic field strengths, the T1 relaxation time of the soft tissue was 382.75–506.48 ms, and T2 was 51.11–70.76 ms. 18F-FDG tracer could be clearly observed in PET imaging. The 3D printed mouse phantom was successfully constructed with tissue-equivalent materials. Our model can be used for CT, MRI, and PET as a standard device for small-animal imaging and quality control.

1. Introduction
Phantoms are widely used for medical imaging performance tests, quality control and assurance testing, medical teaching, and scientific research. Traditional phantoms are machine-manufactured and often only compatible with a single imaging modality. Commercially available phantoms, i.e. Catphan phantom (Ferretti et al. 2016), Gammex phantom (van der Heyden et al. 2017), and AAPM (American Association of Physicists in Medicine) computed tomography (CT) performance phantom (Noh et al. 2014), are designed for CT scanning, while AAPM magnetic resonance imaging (MRI) phantom and American College of Radiology (ACR) MRI phantom...
(Price et al 1990, Chen et al 2004, Bell 2007) are used for MRI quality control. Likewise, the NEMA (National Electrical Manufacturers Association) IEC body phantom (Bradshaw et al 2017) and NEMA positron emission tomography (PET) phantom (Ziegler et al 2015) are mostly used for PET imaging tests (Hattori et al 2013).

Increasingly, research studies and clinical decision-making during treatment rely on multimodality imaging. Therefore, increasing demand for multimodal imaging compatible phantoms exist (Huber et al 2009). Unfortunately, such phantoms are relatively uncommon among current commercial manufacturers and only a few products exist. Those include the PET/CT phantom manufactured by Data Spectrum (Durham, NC, USA), which is used for acceptance testing and routine quality evaluation of PET/CT and single-photon emission computed tomography (SPECT)/CT systems, and the Gillian QA phantom from CIRS (Norfolk, VA, USA), which is used to check alignment and distortion across the entire imaging field on SPECT/CT, PET/CT, and CT/MRI systems.

Most commercial phantoms (as above) are manufactured at very high cost using traditional processing techniques, i.e. molding and casting. But with improvements in 3D printing technology and decrease in 3D printing costs many industries are adopting this new technology for rapid prototyping and product development. For example, in the pharmaceutical industry (Goole and Amighi 2016), construction industry (Domínguez et al 2013), food industry (Lipton et al 2015), and medicine (Marro et al 2016). Compared with traditional techniques, 3D printing utilization has grown dramatically due to the high accuracy, manufacturing efficiency, and ease of customization (Berman 2012). In healthcare, 3D printing has been widely used for precise pre-operative surgical planning (Weber et al 2007) and medical implants to speed up the recovery of patients (Singare et al 2006). In radiation oncology, 3D printing has been used to generate QA phantoms or anthropomorphic phantoms (Niebuhr et al 2016). For example, Niebuhr et al reported a multimodal pelvis phantom composed of various materials compatible with both CT and MRI (Niebuhr et al 2016). Alsabbagh et al fabricated a 3D printed thyroid phantom for dosimetry and image quality evaluation (Alsabbagh et al 2017).

3D printing technology has proven value in medical imaging phantom constructions; however, the use in pre-clinical animal imaging applications is still in development. In radiobiology experiments, mice are often used due to the genetic similarity with humans (Batzoglou et al 2000), and relatively low cost. If radiation will be delivered to a small tumor area, pre-verification of dose delivery accuracy is necessary. This requires a mouse phantom designed with similar anatomy and tissue-equivalency in terms of x-ray attenuation. Additionally, mouse phantoms can be used for quality testing for small-animal imaging. Bieniosek et al developed a 3D printed PET/MR phantom with equivalent function to those commercial phantoms, but lower cost (Bieniosek et al 2015). Perks et al manufactured a 3D printed homogenous mouse phantom replica with inserts for in vitro dosimetry; however, there were no internal anatomic structures (Perks et al 2015). Bentz et al used 3D printing to make a homogeneous mouse phantom with relatively simple geometry for optical imaging (Bentz et al 2016). Welch et al fabricated an anatomic mouse phantom using dosimetrically equivalent materials by stacking multiple machine-made slices (Welch et al 2015), but intricate manufacturing process is complex, expensive, and not scalable.

It has been challenging to identify appropriate materials in 3D printing and achieve tissue equivalency. Commonly used materials such as water glass gel or acrylonitrile butadiene styrene (ABS) are readily available for 3D printing and homogenous phantoms, but are not tissue equivalent for use with CT or MRI. Miller et al printed a PET/CT phantom by powder surface deposition using a liquid binder with radioactive dyes (Miller and Hutchins 2007). This method is complicated and costly, and the production process is affected by the half-life of the radio-tracer. Hunt et al printed a porous cylindrical phantom containing $^{99m}$Tc-pertechnetate or $^{18}$F-FDG for quality control of PET systems (Hunt et al 2009). This traditional cylindrical phantom lacks of human or animal anatomy. Furthermore, radiotracers are often difficult to distribute evenly in printed matrices during manufacturing. Gear et al designed a patient-specific molecular imaging phantom with simplified abdominal organ models, including liver, spleen, kidneys, and multi-positional lesions, but tissue equivalency for those organs was not addressed (Gear et al 2014).

In the present work, various composition ratios of hydrogel and other compounds were evaluated to achieve linear attenuation, tissue-equivalent physical characteristics, and MRI relaxation time using 3D printing technology. Using 3D printing we constructed an anthropomorphic, anatomically-correct mouse phantom with low cost and high accuracy. To our knowledge, this is the first study using 3D printing technology to generate an anatomically accurate, heterogeneous small animal phantom compatible with multimodal imaging platforms.

2. Materials and methods

The heterogeneous mouse phantom is composed of a skeleton, skin shell, and soft tissue materials. Material selection and model designs were based on material characteristics using different imaging platforms.
2.1. Model data acquisition and segmentation

A series of CT images were acquired for a normal healthy mouse with a high-resolution micro-CT (NFR Polaris-G90, NanoFousRay, KOR) with 40 kVp, 0.11 mAs and 0.139 mm slice thickness. Three compartments (head, trunk, and tail) of the whole mouse body were scanned separately due to the length limitation. CT images were imported into Mimics Research 17.0 software to enhance the images by histogram equalization, followed by automatic segmentation of the skeleton and skin shell models, as showed in figures 1 and 2.

2.2. Phantom materials

Fused deposition modeling (FDM) (Stratasys, Minneapolis, MN, USA) and PolyJet (Objet Geometries, Rehovot, Israel) 3D printers were used for phantom fabrication. The main feature of a tissue equivalent material is the ability to match the physical properties with biologic tissue. Materials considered as tissue equivalent should exhibit similar elemental composition, density, and radiation attenuation properties when compared with living tissues (skeleton, muscle, and fat). Radiation attenuation properties, quantified by x-ray attenuation coefficient, CT number (HU), and linear attenuation coefficient (LAC) ($\mu$) of each material, were measured and compared with the mammal using CT scans (Lightspeed-VCT, GE, USA) and Digital Radiography (CXDI-55G, Canon, Japan) with dosimeter (Solid dose 400, German). The LAC energy dependency curves for each material were acquired and plotted against varying tube voltage (kVp) at the same tube current (mAs).

2.2.1. Skeleton model

The skeleton part was fabricated by 3D printing (ObjetConnex 350, Stratasys, Minneapolis, MN, USA) according to the contour segmentation data exported from the CT scan. VERO-WHITE printing material was used to replicate bone, and was wire stocked and processed with photosensitive molding. Special parameter settings were used for bone printing with a density of $1.124 \text{ g cm}^{-3}$, light wavelength at 450 nm, and shore hardness at 90. The production of the skeleton model is shown in figure 1.

2.2.2. Skin shell model

The skin shell was 3D printed using the ABS material (ABS-P430TM XL Model, Ivory) on a FDM 3D printer (Object 50, Stratasys, Minneapolis, MN, USA). The production of the skin shell is shown in figure 2.

2.2.3. Soft tissue equivalent material

In order to achieve CT and MRI compatibility, printed materials used for soft tissue need to be x-ray attenuation equivalent and relaxation time equivalent. Agarose gel has special properties (Derbyshire and Duff 1974, Mitchell et al. 1986), and can be used as a substrate. Micro pearl powder contains CaCO$_3$ and collagen, and can be hydrolyzed into many amino acid and microelement, with similar composition to biological mouse tissues. The NaCl is used as an ion modifier in the gel for conductivity adjustment (Hattori et al. 2013, Cousins 2017).
The x-ray attenuation rate can be improved by adding pearl powder to the gel. The hydrogel was formed by adding agarose into distilled water and mixed with above-mentioned ingredients (agarose, NaCl, and pearl powder), followed by microwave-heating to 100 degree Celsius. A stable geometric shape can be formed after cooling (Mitchell et al 1986, Cousins 2017). $T_2$ relaxation time varies with the ratio of agarose. The magnevist solution is used to vary the $T_1$ relaxation time after being added into the gel. The gel solution can be directly molded and formed after cooling.

After experimentation, we found the ideal composition to be: 1.5% agarose, 1.0% sodium chloride, 1.0% micro-pearl powder, and 0.2 mmol l$^{-1}$ magnevist solution. The attenuation and relaxation time were evaluated for this experimented material.

2.3. Construction of the mouse phantom
Phantom production was divided into forelimbs, trunk, and hindlimbs (see figures 1 and 2 for 3D printed models). These components were attached using hot melted adhesive to form a hollow skin shell (figure 3(A)). The skeleton was inserted and attached to the skin shell matching their relative positions of the anatomical structure. Molten hydrogel was poured into the skin shell and left at room temperature to solidify. Following this the skin and the skeleton were tightly bound by the solidified hydrogel. This process can be repeated by re-heating the phantom in a water bath, pouring off any remaining hydrogel, and refilling the phantom as described above. The characteristics of the skeleton and skin shell do not change with heat-cycling, and can be reused. The hot melted adhesive is polyurethane, which has no effect on CT and MR imaging due to the low density and low x-ray absorptivity or nonmagnetic properties.

2.4. Imaging scans
The assembled mouse phantom was compared with a live mouse using micro-PET/CT (FLEX Triumph, 80 kVp auto-mAs), CT (GE Discovery VCT, 80 kVp/ fixed-mA 20 mAs, Low Dose Brain Protocol), 1.5 T MR (GE, 1.5 T Signa CVI MR), and 3.0 T MR (GE, 3.0 T DISCOVERY MR750). The Micro-PET/CT machine (FLEX Triumph) used in this experiment provides 45–80 kVp and maximum current of 0.5 mAs. Due to visual field limitations with the micro-PET/CT we focused on imaging and comparison of the abdominal region.

For PET imaging, a micro chamber (with a volume of 25 µL) containing 2-deoxy-2-$^{18}$F-fluoro-D-glucose ($^{18}$F-FDG) radiotracer (radiation concentration 1.14 mCi/0.3 ml) was inserted into a pre-designed abdominal cavity in the phantom. Fluorine-18, has a half-life of 109.8 min, and decays by positron emission (0.511 MeV) and electron capture (mean contribution energy of 0.2498 MeV). This simple preliminary imaging test is generally done for PET imaging QA (Buckingham and Gouverneur 2016, Cousins 2017).
2.5. Relaxation time measurement

$T_1$ and $T_2$ relaxation times of various compositions of hydrogel were measured using NMR with various field strength levels: 0.3 T (MacroMR12-150 H-I, 0.3 ± 0.05 T, probe coil diameter 150 mm, master Hertz 12.8 MHz.), 0.5 T (PQ001-20-015 V, 0.5 ± 0.08 T, probe coil diameter 15 mm, resonance frequency 21.29 MHz), 1.0 T (NM-G1, 1.0 ± 0.08 T, magnetic field uniformity ≤20 ppm, resonance frequency: 42 ± 2 MHZ, probe coil diameter 60 mm). For field strengths of 3.0 T, the relaxation times were measured with a GE, 3.0 T DISCOVERY MR750 MRI using a GE workstation v4.6.

Table 1. Physics characteristics of different phantom and mammalian tissues.

| Material            | Density (g cm$^{-3}$) | CT # (HU) | X-ray attenuation coefficient $\mu$ (m$^{-1}$)$^a$ | Melting temperature | Source of data                  |
|---------------------|-----------------------|-----------|---------------------------------|---------------------|---------------------------------|
| VERO-WHITE          | 1.124                 | 130 ± 10  | 54.87–32.03                     | N/A                 | Self-measured                   |
| Mouse skeleton      | N/A                   | 107.91 ± 20 | N/A                             | N/A                 | Self-measured                   |
| Rabbit skeleton     | N/A                   | 145.62 ± 20 | 84.38–37.21                     | N/A                 | Self-measured                   |
| Cartilage           | 1.10                  | 102       | N/A                             | N/A                 | (Goldstone 1990, Wilfried et al 2000) |
| Spongiosa           | 1.14                  | 262 ± 9   | N/A                             | N/A                 | (Goldstone 1990, Wilfried et al 2000) |
| Cortical bone       | 1.92                  | 1524      | N/A                             | N/A                 | (Goldstone 1990, Wilfried et al 2000) |
| Soft tissue gel material | 1.042              | 33 ± 3    | 41.59–24.78                     | 58°C                | Self-measured                   |
| Mouse muscle        | N/A                   | 41 ± 5    | 51.22–30.48                     | N/A                 | Self-measured                   |
| Muscle              | 1.05                  | 45 ± 5    | N/A                             | N/A                 | (Goldstone 1990, Wilfried et al 2000) |
| ABS                 | 0.972                 | −117 ± 3  | 26.10–19.28                     | 260°C               | Self-measured                   |
| Mouse skin          | N/A                   | N/A       | 35.56–24.19                     | N/A                 | Self-measured                   |
| Skin                | 1.09                  | 74 ± 2    | N/A                             | N/A                 | (Goldstone 1990, Wilfried et al 2000) |
| Water               | 1.00                  | 0         | 37.13–22.45                     | 0°C                 | Self-measured                   |

$^a$ With 40–120 kVp x-ray; # with 120 kVp and 100 mAs.
3. Results

3.1. Phantom materials parameters

Physical properties and imaging characteristics were compared for 3D printed products and a live mouse, including the mouse skin and muscle. Because a mouse’s skeleton is too thin to measure x-ray attenuation properties and may result in inaccurate measurements, we tested a rabbit femur in vitro as a surrogate. The

Figure 4. IAC comparisons of experimental materials and mammalian tissue. Water is included as reference. (A) VERO-WHITE versus rabbit skeleton; (B) soft tissue gel material versus mouse muscle; (C) ABS versus mouse skin.

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Table 1 shows the physical density, CT number, x-ray attenuation coefficient, and melting temperature of various test materials. The mean CT number of VERO-WHITE was 130 ± 10 HU compared with mouse bone at 107.91 ± 20 HU, and rabbit bone at 145.62 ± 20 HU. The hydrogel approximation of muscle had a mean value of 33 ± 3 HU, comparable with mouse muscle (41 ± 5 HU), and human muscle (41 ± 5 HU). A comparison of the ABS 3D printed shell to mouse and human skin is shown in Table 1.

The skeletal x-ray attenuation coefficient was measured for the rabbit femur and compared with VERO-WHITE. Figure 4 shows the LAC curves for VERO-WHITE, ABS resin, soft tissue gel material and water. The LAC of bone and VERO-WHITE decrease from 84.38 m\(^{-1}\) to 37.21 m\(^{-1}\) and 54.87 m\(^{-1}\) to 32.03 m\(^{-1}\), with increasing tube voltage from 40 to 120 kVp (Panel (A)). As voltage increases, the x-ray attenuation of VERO-WHITE approaches the value of the rabbit femur. The difference is less than 16.84% between VERO-WHITE and the rabbit femur when tube voltage is greater than 100 kVp. However, as shown in Table 1, the density of VERO-WHITE is between that of human cartilage and spongiosa bone. The CT number of VERO-WHITE is also between CT numbers of cartilage and spongiosa bone. The CT number for the cortical bone (density 1.92 g cm\(^{-3}\)) is higher than 1500 HU. This suggests that VERO-WHITE may only be equivalent to low-density (trabecular) bone. Comparison of hydrogel and mouse muscle is shown in Figure 4 Panel (B). The LAC differences of hydrogel and mammalian muscle decrease with increasing tube voltage: −9.63 m\(^{-1}\) at 40 kVp, −6.55 m\(^{-1}\) at 80 kVp, and −5.70 m\(^{-1}\) at 120 kVp. Panel C shows that LAC decreases for ABS (from 26.10 m\(^{-1}\) to 19.28 m\(^{-1}\)) and skin (from 35.56 m\(^{-1}\) to 24.19 m\(^{-1}\)) as the tube voltage increases from 40–120 kVp. CT numbers and attenuation coefficients of water were measured under the same conditions (marked in red in figure 4) and used as a reference for comparison.

### 3.2. Multimodality imaging tests

The micro-PET/CT images of the mouse phantom are shown in Figure 5. The tracer area of the container can be clearly observed. CT images of the mouse phantom acquired at 80 kVp/250 mAs and 80 kVp/310 mAs are shown in Figure 6.
in figures 6(A) and (B). Good contrast between bones and hydrogel can be observed on both scans, and the latter shows reduced noise due to higher tube current.

MR images of mouse phantom at 1.5 T are shown in figure 7. We selected $T_1$-SE (TR: 383 ms, TE: 9 ms, FOV: 160 mm, Slice thickness: 5.0 mm, Matrix: 256 × 192), $T_2$-FRFSE (TR: 2367 ms, TE: 108 ms, FOV: 160 mm, Slice thickness: 5.0 mm, Matrix: 256 × 192) and 3D FLESTA-C (TR: 5.8 ms, TE: 2 ms, FOV: 160 mm, Slice thickness: 0.5 mm, Matrix: 256 × 160) sequences to scan the trunk of the phantom. The trunk abdominal bone shows no signal on MR images. The simulated soft tissue presents low signal on $T_1$ and $T_2$ images, and signal intensity and imaging performance are similar to that of muscle and cartilage on MR scans. The contrast and resolution of the image are improved with the 3D FLESTA-C sequence. With the maximum intensity projection (MIP, figure 7 Panel (g)), the partial structure of the mouse phantom is clearly seen. MR scans of the phantom at 3.0 T (without the micro-container inserted) are shown in figure 8. Imaging sequences include $T_1$WI-SE (TR: 15.00 ms, TE: 2.32 ms, FOV: 160 mm, Slice thickness: 3.0 mm, Matrix: 256 × 256), $T_2$WI-SE (TR: 1000 ms, TE: 13.8 ms, FOV: 160 mm, Slice thickness: 3.0 mm, Matrix: 256 × 256), and $T_2$-STAR (TR: 607 ms, TE: 4.36 ms, FOV: 160 mm, Slice thickness: 3.0 mm, Matrix: 256 × 256). The $T_1$ and $T_2$ relaxation times of the hydrogel are shown in table 2. The relaxation times of the simulated soft tissue increase with magnetic field intensity and approach the relaxation time of the muscle under the same condition. $T_1$ and $T_2$ relaxation times are 506.48 ms and 70.76 ms at 3.0 T scanner. Table 3 shows published $T_1$ and $T_2$ relaxation times for mouse and human tissues for comparison.

4. Discussion

There are several advantages to building a mouse phantom with 3D printing technology. Material consumption is reduced; there is excellent spatial accuracy for printed materials; and generation of highly complex geometries is possible (Edgar and Tint 2015). Using 3D printing, anthropomorphic animal phantoms or organ models of any type or shape can be efficiently and cost-effectively manufactured (Welch et al 2015). However, previous studies mostly focused on homogeneous designs for single imaging modality use, and some lack the tissue equivalency necessary for use with multimodal imaging platforms (Bieniosek et al 2015, Bentz et al 2016). Appropriate materials selection can simulate living soft tissue both in CT and MRI; however, current commercially available
3D printing materials cannot meet these demands. Casting is still the predominant manufacturing method for generating imaging animal phantoms (Huber et al 2009). This method is also applied for manufacture of soft tissue organ models, with 3D printed hollow shell filled with tissue equivalent gels. We found that CT numbers and MRI relaxation times can be optimized to mimic living tissue by changing the mass fraction of micrometer-sized pearl powder and the amount of gadolinium agent (within a certain range) of the hydrogel.

In the present study, we chose soft tissue materials according to the characteristics of the simulated tissues, including density, radiation characteristic, LAC, and relaxation times. It is important to note that there is no substitute capable of exactly matching the corresponding mammalian tissue reference. By adjusting the content of the high Z elements added in the substitute hydrogel material, the $\mu/\rho$ and $\mu$ values of the material can be close, or equivalent to, the biological reference tissue (Goldstone 1990). Common radiation tissue-equivalent materials are gellan gum (Renvall 2009, Hattori et al 2013) and agarose (Derbyshire and Duff 1974, Yohannes et al 2012, Hattori et al 2013). Considering agarose’s similarity to the mammal tissue in terms of the attenuation characteristics, hydrogen proton composites, physical characteristic, and simple preparation process (Yohannes et al
obtained the ideal proportion of relaxation time equivalent material of the soft tissue. Hollis (1973) measured relaxation times were studied by adding and adjusting the amount of Gadopentetate dimeglumine. We studied material is within the uncertainty range of the living mouth skeletal muscle. Damadian et al. (1974) made measurements on a pulsed NMR spectrometer at 24 MHz analyse human breast tissues (Medina et al. 1975).

The relaxation time of pure agarose gel (similar to the relaxation time of water) can be varied when adding cupric ion or Gd ion (Hattori et al. 2013). In MRI, magnetivist solution is a common paramagnetic contrast agent, and can shorten the $T_1$ relaxation time (Cheng et al. 2006, Renvall 2009). The agarose gel is often used in the fabrication of MRI phantom, and the relaxation time of the gel can be varied by adding moderate amount of paramagnetic materials (Cheng et al. 2006, Renvall 2009, Hattori et al. 2013). On the basis of the CT equivalent gel, the equivalent relaxation times were studied by adding and adjusting the amount of Gadopentetate dimeglumine. We obtained the ideal proportion of relaxation time equivalent material of the soft tissue. Hollis et al. (1973) measured $T_1$ relaxation time ($\pm$SE) of mouse skeletal muscle using a pulsed NMR spectrometer at 24 MHz, with a value of 411 ± 40 ms as shown in table 2. The mean $T_1$ value for soft tissue gels done at a resonance frequency of 21.29 MHz was 391.05 ± 32.76 ms. In table 3, the $T_1$ relaxation time of the mouse skeletal muscle is reported to be 411 ± 40 ms, where rat rectus muscle is shown to be slightly higher (538 ± 15 ms). The $T_1$ relaxation time of our studied material is within the uncertainty range of the living mouth skeletal muscle. Damadian et al reported the $T_2$ relaxation time of the rat rectus muscle with a value of 55 ± 5 ms (Damadian 1971). The mean $T_2$ value for the agarose gels obtained at a resonance frequency of 21.29 MHz was 61.20 ± 5.32 ms. Although lacking solid data from literature regarding the $T_2$ relaxation time for the mouse skeletal muscle for comparison, the $T_2$ value is considered to be in an comparable range. Overall, it has been demonstrated that our experimented soft tissue material is within the range of the reported normal mice and human muscles in terms of the $T_1$ and $T_2$ relaxation times.

Our 3D printed mouse phantom resembles the real mammal mouse in terms of the body external contour, skeleton shape, CT numbers of the bone and soft tissue, and relaxation times. The technology we employed, and our approach, can be used for other studies of radiation therapy, i.e. dosimetric verification and measurements. Through advanced material selection and use of 3D printing technology we were able to fabricate an anthropomorphic tissue-equivalent mouse phantom that can be used for animal imaging and studies with multiple imaging platforms. Other previously developed multimodal imaging phantoms (Chen and Shih 2013) can be also re-constructed using this method with decreased cost and increased flexibility. Our present method supports 3D printing equivalent human skeleton and skin models in combined with soft tissue and organ casting using gels that are similar to human tissue characteristics, including physical properties, radiation equivalence, and equivalent relaxation time.

Our future work will focus on further optimization of substitute soft tissue materials to better simulate living tissue characteristics. Additionally, to improve the similarity of modeled versus biologic PET scanner QA, we hope to incorporate radioisotopes into the hydrogel (18F or 11C) in their molten state to obtain a more realistic PET activity distribution (Markiewicz et al. 2011).

5. Conclusions

We showed that soft tissue equivalent materials can be used with 3D printing technology to generate an anthropomorphic mouse phantom which improves upon current commercial phantom production techniques and modeling. Multimodality imaging (CT, MRI, and PET) confirms the tissue equivalent properties of substitute materials.
soft-tissue materials compared with in vivo biologic systems in terms of physical characteristics, x-ray attenuation characteristics, and relaxation times. This phantom can be applied in small animal research imaging studies.

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