The ADAM-pelvis phantom—an anthropomorphic, deformable and multimodal phantom for MRgRT

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Abstract

Applicability and accuracy of the rapidly developing tools and workflows for image-guided radiotherapy need to be validated under realistic treatment-like conditions. We present the construction of the ADAM-pelvis phantom, an anthropomorphic, deformable and multimodal (CT and MRI) phantom of the male pelvis. The phantom covers patient-like uncertainties in image-guided radiotherapy workflows including imaging artifacts for the special case of the human anatomy as well as organ motion.

Principles and methods were further improved from previous work. The phantom includes surrogates for muscle tissue, adipose, inner and outer bone, as well as deformable silicone organs. Anthropomorphic shapes are realized with 3D-printing techniques for the bone and the construction of the hollow silicone organ shells. Organs are constructed from patient image segmentation and further guided by reported deformation models. Imaging markers and pockets for dosimeters are included in the organ shells.

The improved phantom surrogates match imaging characteristics in MRI (T1 and T2 relaxation time) and CT (Hounsfield units) of human tissues. The surrogates are suited for long term use (several months) of the phantom. Previously reported artifacts of the muscle surrogate were avoided by improved composition of the used agarose gel. Interfractional organ motion is successfully realized for the water filled bladder and the air filled rectum and showed to be reproducible with deviation below 1 mm. Volume variations of both induce displacement, rotation and deformation of the prostate.

We present solutions for the construction of an anthropomorphic phantom suitable for MRI and CT imaging including deformable organs. The developed concepts of phantom surrogates and construction techniques were successfully applied in building the ADAM-pelvis phantom and can as well be adopted for other anthropomorphic phantoms. The presented phantom allows for the systematic and controlled investigation of image-guided radiotherapy workflows in presence of organ motion.

1. Introduction

With advances in image guided radiotherapy (IgRT) imaging artifacts, organ motion and signal intensity uncertainties create challenges in all modalities used in the process of IgRT. To be able to exploit the newly gained advantages of daily imaging, validation of the now available data, tools and workflows is required (Jaffray \textit{et al} 2010). Especially for the dose accumulation in adaptive radiotherapy, or, for the dedicated case of MR-guided
radiotherapy, where questions arise about the spatial accuracy of MR images as well as about the quality of synthetic CTs based on MRI.

In patients, all these sources of uncertainty occur simultaneously, which strongly hampers the development and quality assurance of new tools that address the individual problems. Single, isolated questions can be addressed in phantoms where ground truth can be available. However, many of the occurring effects originate from the characteristic features of the human body and can therefore hardly be imitated by abstract phantoms. Anthropomorphic phantoms are needed that enable both the separate and combined investigation of uncertainties in patient-like setups and treatment scenarios. Since image guidance focuses on the reduction of uncertainties due to organ motion, the possibility to imitate moving and deforming organs is an essential feature that needs to be covered.

Here, the ADAM-pelvis phantom, an anthropomorphic, deformable and multimodal phantom realized for the male pelvic region is presented. The aim is to introduce an anthropomorphic phantom suitable for imaging in (dual energy) CT as well as MRI and for use in adaptive IGRT scenarios, including interfractional organ motion. This paper focuses on the detailed instruction on the overall construction of the phantom and its compartments and the adaptations to the already presented tissue surrogates while pointing out the improvements to previous work (Niebuhr 2016).

2. Materials and methods

This paper builds on and improves methods presented in Niebuhr et al (2016) and will therefore focus on detailed descriptions of the changes and further developed aspects building on the prior basis.

2.1. Overview and construction

Figure 1 shows the complete setup of the ADAM-pelvis phantom ready for measurements (a). The outer phantom construction is based on the dimensions of a realistic patient (elliptical cross-sectional area of 370 × 220 mm) and made from polymethylmethacrylat (PMMA, CT values: 129 HU ± 5 HU at 120 kV tube voltage). A tube enters through the upper lid that connects the bladder to the outside, enabling manual variation of the bladder volume with a syringe.

The 3D-printed bone model (figure 1(b)) is fixated inside the PMMA case (section 2.2.). A total volume of 8 l of muscle surrogate (section 2.3.) encloses the bone, providing central space for organ placement (figure 1(c)). The remaining phantom volume is filled with oil as adipose tissue surrogate in accordance to the human anatomy, where the intraabdominal adipose tissue serves as a spacer between the organs (section 2.4.).

To avoid overpressure inside the phantom when increasing the organ volumes, the displaced oil can flow into an extra bag which is attached to the upper lid (figure 1(a)). This is also used to remove oil before handling the organs (e.g. for changing detectors) inside the phantom. A valve on top ensures an air-enclosure-free filling of the phantom.

One further opening in the outer case is embedded at the bottom of the phantom (caudal) with direct access to the interior of the rectum. Here, a silicone balloon connected to a syringe can be inserted from the outside to expand the rectum diameter (from 20 to 30 mm) by increasing the balloon volume with either liquid or air.

For measurements, the phantom is placed on a 3D-printed mount (figure 1(a)). The phantom case and the mount are manufactured such as to allow controlled rotation of 2° or 4°, to simulate treatment setup errors in a controllable manner.

2.2. Outer bone

To achieve stable outer bone CT values (1600 HU, Schneider et al 2000), strong gypsum (figure 1(b)), was used as surrogate on top of the previously described 3D printed hollow bone (Niebuhr et al 2016). A thin layer of clear lacquer was added for additional protection. By not absorbing water, the strong gypsum proved to be more resilient over time than the previously used gypsum bandages. It also showed better agreement with outer bone CT values. The inner part of the bone surrogate remained unchanged and comprised of a mixture of Vaseline and K2HPO4 to simulate bone marrow.

2.3. Muscle tissue surrogate

As previously reported, agarose gels (for T2-adaptation) doped with a gadolinium (Gd)-based contrast agent (MultiHance, 0.5 M, for T1-adaptation) and sodium fluoride (NaF, adaptation of CT values) successfully provide a wide range of possible tissues to be simulated in MRI and CT by varying the ingredient concentrations accordingly.

Reported in Niebuhr et al (2016), a strong signal fall off was visible in the central phantom on MR images, possibly due to high salt concentration of NaF causing a high conductivity of the material. Similar effects are described by Merkle and Dale (2006) as standing wave and conductivity effects. The conductivity of the former
used agarose gel was calculated to be in the order of \( \lambda (4\ w\%\ NaF) \approx 9\ S\ m^{-1} \) only considering the salt concentration. Gabriel et al (1996) report a conductivity of muscle tissue of no more than 1 S m\(^{-1}\). Reducing the salt concentration as necessary requires an increase in the concentration of the other components to still facilitate muscle tissue mimicking. By lowering the salt concentration as much as possible while still achieving muscle-like CT values, as an appropriate compromise, a water based gel with 3.5 w\% agarose (Agarose HEEO Ultra-Quality, Carl Roth GmbH), 1 w\% NaF and 0.025 w\% MultiHance solution for the simulation of muscle tissue is tested inside the finished phantom setup (table 1).

The muscle gel is cast around the rigid pelvic bone inside the cylinder, only leaving out the organ cavity in the center. Figure 1(c) shows the open phantom in preparation right after hardening of the gel, still including the black spacer. It is later removed when the gel is completely cooled down. Concentrations of the prostate and tumor surrogate are directly applied from the previous work (table 1).

2.4. Adipose tissue surrogate

As an adipose tissue surrogate, various vegetable oils were successfully tested in Niebuhr et al (2016), showing only a small difference in the radiological properties of different samples. The olive oil of the previous version was replaced by peanut oil. Reasons were its similar imaging properties with the advantages of its very light and translucent appearance (for operating inside the phantom) as well as its high oxidation stability.

2.5. Production of silicone organ models

New organ models were developed of the prostate, bladder and rectum with improved anthropomorphic shapes and deformation properties (figure 2). Two further tasks were the implementation of marker points and detector bags into the surface of the deformable organ shells. A detailed description of the production of the different organ models can be found in the supplementary material (stacks.iop.org/PMB/64/04NT05/mmedia).

A 3D rendering of the organ models and a picture of the produced silicone shapes is depicted in figure 2. Analogously to the human anatomy, the prostate model encloses the urethra represented by a tube at the bottom of the bladder (g), so that the prostate is attached to the bladder but can move in all spatial directions and can also undergo a twisting motion (Huang et al 2015). The bladder is attached to the bone at the two upper tubes (representing the entering urinary tracts from both kidneys (figure 2(a))) and at the anterior side while still enabling motion in all spatial directions. One of the upper tubes is connected to a syringe outside of the phantom to allow manual interfractional variation of the bladder volume (see figure 1).

The rectum model is not attached to the other organs. It is fixated inside the phantom at the bottom of the plastic cylinder with an accessible opening. At the posterior side, the rectum is fixated to the pelvic bone at the coccyx.
Table 1. Results for CT numbers (CT #) in Hounsfield units (HU) and MRI relaxation times (T1 and T2 at 1.5 T in ms) for tissue surrogates used in the ADAM-pelvis phantom compared with reported reference tissue values in literature.

| Reference | Tissue | CT # (HU) | T1 (ms) | T2 (ms) | Phantom Surrogate | CT # (HU) | T1 (ms) | T2 (ms) |
|-----------|--------|-----------|---------|---------|------------------|-----------|---------|---------|
|           | Muscle | 40–44<sup>b</sup> | 856 ± 61<sup>e</sup> | 27 ± 8<sup>c</sup> | Agarose 3.5 w%, NaF 1 w%, MH 0.025 w% | 26 ± 3 | 886 ± 4 | 20 ± 1 |
|           | Muscle | 54 ± 7<sup>d</sup> | 47 ± 13<sup>e</sup> | 26 ± 3 | Strong gypsum | 1845 ± 75 | No signal | No signal |
|           | cortical bone | 1524<sup>b</sup> (rho 1.78, Z<sub>eff</sub> 13.4)<sup>a</sup> | No signal | No signal | Peanut oil | −102 ± 23 | 283 ± 25 | 110 ± 3 |
|           | Adipose | (−95)–(−55)<sup>b</sup> | 343 ± 34<sup>e</sup> | 58 ± 4<sup>c</sup> | Agarose 1 w%, NaF 4 w%, MH 0.013 w%<sup>f</sup> | 41 ± 3 | 1407 ± 42 | 94 ± 2 |
|           | Prostate | 34<sup>b</sup> | 1317 ± 85<sup>b</sup> | 88 ± 0<sup>c</sup> | Agarose 2 w%, NaF 4 w%, MH 0.010 w%<sup>f</sup> | 41 ± 3 | 1416 ± 53 | 69 ± 2 |
|           | Prostate carcinoma | Approx. same as prostate<sup>d</sup> | Approx. same as prostate<sup>d</sup> | Shorter than prostate<sup>f</sup> | Agarose 2 w%, NaF 4 w%, MH 0.010 w%<sup>f</sup> | 41 ± 3 | 1416 ± 53 | 69 ± 2 |
|           | Inner bone (yellow/red marrow) | 140 ± 170<sup>c</sup> | 549 ± 52<sup>e</sup> | 49 ± 8<sup>c</sup> | Vaseline 75 w%, K$_2$HPO$_4$ 25 w%<sup>b</sup> | 134 ± 38 | 169 ± 27 | 49 ± 1 |

<sup>a</sup> CT numbers from calculation based on tissue composition in Woodard and White (1986).
<sup>b</sup> Schneider et al (2000).
<sup>c</sup> Bazelaire et al (2004).
<sup>d</sup> CT numbers measured in the underlying patient data at 120 kVp.
<sup>e</sup> Bottomley et al (1984).
<sup>f</sup> Villeirs and De Meerleer (2007).
<sup>g</sup> Same composition as used in Niebuhr et al (2016).
Three different types of silicone are used for the production of the organ models. The translucent TFC silicone (Trollfactory) type 13 shows high elasticity with a shore hardness of 00 and is closest to soft tissue in its radiological properties (CT number (at 120 kVp) = (169 ± 5) HU, T1 = (807 ± 15 ms), T2 = (246 ± 23) ms, see also table 1) compared to stiffer silicone types (see the following). Varying the thickness of the silicone allows more targeted deformation towards the main desired expansion directions. It is therefore chosen as main casting material for the bladder and the prostate model (figures 2(d) and (f)).

Orange Neukasil RTV22 with shore hardness of 22 does not show the desired elasticity for the bladder but provides stability in critical regions. It is therefore used for reinforcements where the translucent silicone is not stable enough: at the bladder were tubes are fixated (figure 2(b)) and for the integration of detector bags on the caudal side of the bladder (figure 2(e)). It shows high values in CT with 218 HU (T1 = (782 ± 11) ms, T2 = (123 ± 6) ms). The orange silicone is also used for the construction of the complete model of the rectum (figure 2(h)) since in this area stability is a more important issue in the construction and softer silicone would lead to uncontrolled deformation and rips.

Yellow TFC silicone type 2-1 (Trollfactory) is used as marker points on the organ surface (figure 2(c)) of bladder and prostate and will deform with the organ shell. With an even higher shore hardness of 35, the yellow silicone shows the strongest contrast to the surrounding translucent silicone in all modalities. Its CT values are (355 ± 8) HU and MRI relaxation times were measured to be T1 = (581 ± 19) ms and T2 = (35 ± 2) ms.

The shapes of the silicone models are based on organ delineations from patient data. In the case of the prostate and the rectum, a single patient CT-dataset served as template for the construction of the synthetic organs. Organ segmentation is performed manually in the software MITK (Nolden et al 2013) using 3D-interpolation methods. The converted STL file is imported into the software Geomagic Freeform and Autodesk Inventor for processing the tessellated surface into casting moulds.

For the bladder, not only the shape but also the characteristics of its deformation are considered by using a bladder deformation model described by Stoll et al (2016). The model is used to retrieve information about the shape of the bladder in different filling stages. According to the delineation and the biomechanical model, not only the shape of the bladder is designed but the construction also included regions with higher wall thickness of the silicone in order to give a preferred direction of expansion. It is found that the main deformation direction was the crano-caudal direction with limited expansion to the sides of the bladder (left-right). Therefore, for example higher wall thickness is implemented at the sides of the bladder model (left-right and anterior–posterior 4 mm wall thickness) and thinner wall thickness in cranio-caudal direction (3 mm). It was found that a minimum of 3 mm wall thickness is necessary to securely include silicone marker points into the bladder wall.

2.6. Imaging
All CT measurements are performed with a Siemens Somatom Definition Flash scanner at 120 kVp with a slice thickness of 2 mm. MRI scans are performed with a Siemens Magnetom Symphony 1.5 T scanner. For the measurement of the absolute T1-relaxation time, an inversion recovery method with a turbo FLASH (fast low angle shot) sequence is used (slice thickness 10 mm, TI = 300–9000 ms in 21 steps, TE = 1.26 ms, TR = 10000 ms, flip angle 9°). T2 is measured using a multi spin-echo sequence with TE = 9.2 * (1:32) ms (slice thickness = 10 mm, TR = 3000 ms). Morphological MR imaging of the phantom is done with a standard imaging sequence used for patients of the pelvis at our institute. The sequence is a T2-weighted SPACE sequence.

Figure 2. (Left) 3D-rendering of the models of bladder, prostate and rectum as used for the construction of casting moulds. (Right) Photo of the produced silicone organ models.
with a voxel size of $1.1 \times 1.1 \times 1.0$ mm ($\text{FoV} = 500$ mm, $\text{TR} = 2000$ ms, $\text{TE} = 124$ ms, bandwidth 657 Hz/px, flip angle $150^\circ$, using built in Siemens 3D distortion correction). Visualization of the images for display in this work is done using the software MITK (Nolden et al 2013).

3. Results

A direct comparison of a CT and an MRI scan of the phantom and a patient is presented in figure 3 applying identical imaging protocols in each modality.

Surrogate tissue pixel values and contrasts of the phantom are well comparable to in vivo data in most areas for both CT and MRI. No non-physiological signal fall-off is visible in the central region as was reported in Niebuhr et al (2016). Important anthropomorphic properties are implemented. Marker points are visible on the organ surfaces in the MRI scan (arrows). Organ shells show higher signal intensity in CT than in the patient case.

3.1. Material properties

The summary of all results of the radiological material properties compared to human tissues can be found in table 1. Furthermore, a comparison of the signal distributions in CT and MRI of the current ADAM pelvis phantom setup, the former phantom setup and a number of measured distributions of patients is given in figures 5(a) and (b).

3.1.1. Cortical bone

Strong gypsum covers a CT value range of up to 2200 HU with a mean value of 1845 HU. Figure 4 shows a comparison of the phantom bones of the former and current setup, and a patient bone in a colored surface representation of CT numbers in HU. The currently used phantom bone covered in strong gypsum shows a similar distribution of high and low CT number areas compared with the patient bone, achieved by applying different gypsum thicknesses, yet already exceeding the range of values found in the patient. No decrease in CT numbers is visible for the strong gypsum in contrast to the gypsum bandages. No signal is visible on the bone surface in the MRI scans (figure 3).

3.1.2. Muscle tissue

The muscle surrogate shows good stability inside the phantom forming the organ cavity. Mean CT values for muscle tissue are 26 HU represented by the highest (green) peak in figure 5(a). Compared to values found for patients and the former phantom setup the according green peak in figure 5(a) is shifted towards lower values. For the intensity distribution in MRI (figure 5(b)) the lower range is covered by the muscle tissue, comparable to human tissue and similar to the former setup.

3.1.3. Adipose tissue

The radiological properties measured for peanut oil are in agreement with previously reported values (Niebuhr et al 2016) and are close to those of intraabdominal fat in the human body in CT and MRI (table 1). Figure 5(a) shows a peak in the lower range of CT values (around $-100$ HU) that corresponds to fat. Figure 5(b) shows the...
signal intensity in MRI where the oil corresponds to a peak in the mid intensity range. The abundance of high intensities is generally higher for the current ADAM pelvis phantom setup than for the former setup.

### 3.1.4. Silicone organs

The silicone organ shells appear dark in MRI, while in CT they are more clearly visible as hyperintense boundaries. The silicone organs as well as the PMMA case lead to an additional peak in the range of 120 HU in figure 5(a) which is not present in the given distributions in patients. The contrast between the translucent silicone of the bladder and prostate wall and the yellow marker points is high enough in MRI to be distinguishable, however, in CT no marker points can be extracted.

### 3.1.5. Stability over time

The radiological properties were quantitatively investigated inside the filled phantom over a period of 2 months. For the muscle gel, an increase in the T2 relaxation time of 6 ms was observed (20 to 26 ms). T1 shortened by approximately 20 ms (886 to 866 ms). CT values were stable in the range of error but showed a small tendency towards higher values (3 HU within 2 months). No relevant changes of the properties of the gel inside the prostate were observed. Fat was stable in its CT numbers as well as the T2 relaxation time, while T1 shortened by about 60 ms. In contrast to the former phantom cortical bone, no changes to the attenuation properties were observed for strong gypsum in contrast to gypsum bandages. Furthermore, in the use of the phantom over a period of 6 months and longer, no qualitative changes in the signal intensities were observed. The preservation of the muscle and prostate gel, especially protection against drying out, is given by the covering oil.

The phantom has then been in frequent use for various applications over a period of 2.5 years. Soft tubes (e.g. water access to the bladder) had to be replaced after periods of about 9–12 months because the surrounding oil removes plasticizers from the used plastic materials and tubes became less flexible and brittle. There was no change in flexibility of the silicone organs. Measurements of the tissue surrogate properties inside the phantom showed only marginal changes in HUs for all materials (below 5 HU difference). For the used oil, T1 remained stable while T2 increased by 40 ms from 108 to 148 ms. The bone marrow surrogate showed a slight decrease in T1 of 30 ms and an increase in T2 from 49 to 63 ms. There was an increase of T2 for the muscle gel (from 20 to 43 ms) and the highest overall difference was measured for T1 of muscle increasing from 886 to 1460 ms. Both changes in T1 and T2 might be due to the Gd-contrast agent that was used to decrease T1 which might have settled towards the phantom bottom through the net-like structure of the agarose-gel. For CT imaging and T2-weighted MRI that was performed regularly over the course of 2.5 years, no visible alterations in the imaging contrasts were seen.
3.2. Synthetic organs and motion

Shapes of the synthetic organs are well comparable to those of the human anatomy (figure 3). Repeated CT measurements of the six tested anatomical states showed a high level of reproducibility with accuracy of 1 mm or better in the organ positions.

Four different morphological states of the phantom organs inside the ADAM pelvis phantom are presented in figure 6, showing the minimum and maximum expansion of bladder and rectum. Maximum achievable bladder expansion was tested from a base level of 150 ml to a medium state of 250 ml and to a maximum of 350 ml. Lowering the volume introduces a dent on the top of the bladder. The bladder expanded in all spatial directions with a preferred direction towards the superior side. The shape of the expanded bladder showed to be reproducible without signs of wearing out.

Both bladder and rectal expansion introduced motion of the prostate, which represents the target volume. While expanding the bladder mainly shifts the prostate in the caudal direction, expansion of the rectum introduces translation towards the anterior side as well as a rotation. The influence of the bladder expansion is strongest when increasing the bladder from the ground level of 150 ml to the medium level of 250 ml with a maximum point position difference of 5 mm and a prostate center shift of 4 mm mainly in caudal direction. Increasing the bladder volume to the maximum level of 350 ml does not increase the central prostate position shift any more while single marker points show an additional translation of 1 mm compared to the medium volume stage (twisting). The induced motion of the prostate shows a tilt of the organ towards the inferior direction of the anterior side of the prostate.

Figure 5. Distribution of signal intensities in (a) CT (given in HU) and (b) MRI (for a T2 SPC sequence) for the ADAM-pelvis phantom setup (green), the former phantom setup (gray) as presented in Niebuhr et al (2016), as well as example distributions of the pelvic area of eight patients (Bostel et al 2014) (blue). PV: partial volume, represented by mixed voxels of muscle and adipose tissue.
The influence of the rectal filling stage on the position shift of the prostate is generally lower than that of the bladder. Expanding the rectal ball volume by 30 ml (maximum) increased the rectal diameter by 10 mm. This introduced a maximum shift of the center of the prostate of 2 mm towards the anterior side. A maximum marker point shift of 3 mm is observed due to a twisting motion of the prostate.

4. Discussion

The ADAM-pelvis phantom combines multimodal applicability in imaging (CT, MRI) and the simulation of treatment scenarios in adaptive radiotherapy by enabling organ deformation. The specific focus is put on the realistic interplay of motion between different organs as occurring in patients. The combination of all these properties is not reported for other multimodal phantoms in literature, especially not for the pelvic region. A number of reports cover the development of deformable lung phantoms (e.g. Chang et al (2010) and Cherpark et al (2011)) based on the underlying importance of motion management in the thorax. While these phantoms address the specific question of lung motion, this does not include the effect of motion on other organs than the lung or the included tumor. Other phantoms are specifically designed to simulate realistic motion characteristics of a set of organs but are focused on the deformation caused by needle-insertion rather than organ-induced motion patterns (Hungr et al 2012, Ruschin et al 2016). The ADAM-pelvis phantom simulates the whole pelvis of a patient, thus, realizing a complete patient setup in the relevant region. Based on the phantom dimensions and materials, realistic simulation of imaging artifacts, such as beam hardening in CT or geometric distortions in MRI are included in a patient-like manner. This setup also facilitates the use of the phantom for radiotherapy applications, where the total cross section of the patient’s region of interest is important. These aspects cannot be realized with phantoms that only cover a distinct volume of the body, like a single organ or a set of organs of interest as for example presented by Chmarra et al (2013) for liver or by Wang et al (2005) for the inner pelvis although these phantoms realize a variety of solutions for multimodal and deformable setups.

4.1. Material properties

The presented materials used in the ADAM-pelvis phantom agree well with reported and measured values for human tissues in their imaging properties in MRI and CT. As reported in the previous work, the chosen materials have been tested for the use in photon as well as particle therapy by measuring and adjusting the electron density and effective atomic number to fit tissue measurements using dual energy CT. By using the same set of materials, the presented phantom is well suitable to mimic the photon attenuation as well as particle interactions found in human tissue.

4.1.1. Cortical bone

Strong gypsum is found to be a well-suited surrogate for cortical bone representation showing the desired high and stable attenuation properties without producing any signal in MRI. In figure 4 it can be seen that in some areas the phantom bone shows higher CT values than found in the patient. The desired attenuation can however be further adapted by changing the thickness of the gypsum layer.

4.1.2. Muscle tissue

The T1- and T2-relaxation times fits the reported literature values (table 1) and the signal intensity distribution in MRI is well comparable to that of patients (figure 5(b)). Thus, with the presented combination of materials, a multimodal use of the phantom is realized and justifies the trade-off between CT values and signal intensity in MRI. Originating from this trade-off, the results for the CT values of the muscle gel are lower than those reported for muscle tissue in literature and the former phantom setup (figure 5(a)). However, the former distribution misses the amplitude in the range between soft and fatty tissue. This is mainly due to the fact that in the phantom, there is a clear separation between fat and muscle tissue with little transition area between the two surrogates. In a patient, muscle is usually streaked with fatty areas which result in a high abundance of voxels showing partial volume (PV) effects (muscle/fat). The CT values of the here presented muscle surrogate cover a larger area of the intermediate PV-range between fat and muscle representing a muscle tissue CT-value closer to a mixture with fatty streaks. Nevertheless, the presented phantom shows contrasts between tissues that are well comparable to patient scans. Additionally, dosimetric properties of the tissues are well adapted to human tissues in the way that the match in CT values is achieved by matching both the electron density and the effective atomic number rather than focusing on the CT number alone as reported in more detail in Niebuhr et al (2016). Thus, the presented phantom serves well as an anthropomorphic model for CT as well as radiotherapy applications.

The minor occurring changes in the radiological properties of the soft tissue surrogates over a period of 2 months do not alter the use of the phantom for imaging. It was not found to be necessary to add preservatives to the gel used in the phantom setup (as for example reported in D’Souza et al (2001), Hattori et al (2013) and Huber et al (2009)) because no decay or alteration of its properties was observed due to the preserving effect of
the surrounding oil, which was found to be essential for the long-term stability of the phantom. For long-term use, stronger changes only occurred for the T1 relaxation time of the muscle gel, suggesting that it should be replaced after usage of a maximum of 2 years to ensure tissue-like properties in MRI.

4.1.3. Adipose tissue
The radiological properties of the fat surrogate show good agreement with adipose tissue in all modalities. The handling inside the phantom and quality assurance from the outside is greatly simplified because of the good transparency of the oil which is especially important in the large volume of the presented phantom. Its stability over time easily exceeds 6 months with proper handling (use of gloves and sealing when not in use).

4.1.4. Silicone organs and plastics
Handling and deformation properties of the described silicone types showed to be very well applicable for the production and use of deformable and expandable anthropomorphic organ shells. The use of stronger silicones is essential to ensure the stability of the organs at critical points.

The low signal intensity in T2-weighted MRI of all used silicone types matches well with the observable low signal of organ walls (figure 3). The contrast between the translucent and the yellow silicone in the bladder and prostate wall enables the differentiation of the single marker points as hypointense spots. These can be used in manual registration and their applicability in automated registration is to be tested for future studies. Since the markers are only visible in MRI, a CT–CT registration might be verified using the MRI scans. In CT imaging the organ shells appear brighter in the phantom than in patients, therefore automatic segmentation and registration might be simplified compared to patient images.

An additional peak in the range of 100–150 HU can be seen in the CT number spectrum of the phantom that does not occur for patients (figure 5(a)). This peak is caused by silicone, the plastic of the printed bone case and to a large part by the PMMA shell of the phantom. It would be beneficial to use a material with lower absorption properties, especially for the phantom case. However, to the author’s knowledge there is no alternative material available that shows the same stability, handling and processing quality to be used to fabricate the case of a phantom other than PMMA.
4.2. Organ motion
The simulated motion of the prostate feasible within the ADAM-pelvis phantom is realistic in terms of manner, direction and amplitude compared to patient data. The displacement of the phantom-prostate was preferably directed in the caudal direction in case of increased bladder filling (up to 5 mm) and in the anterior–posterior direction for rectal volume changes (up to 3 mm). Studies on prostate motion agree in the fact that it is mainly directed in the caudo-cranial and anterior–posterior direction (Britton et al 2005, Litzenberg et al 2006, Kotte et al 2007, Langen et al 2012). Motion amplitude of the prostate is reported by Kotte et al (2007) to be outside of 2 mm in 66% of studied fractions and outside 3 mm in 28% of fractions with highest probability in the caudo-cranial direction. Mean displacement values found by Britton et al (2005) are in the range of 3 mm in cranio-caudal and anterior–posterior direction. Furthermore, a rotation and tilting of the prostate inside the ADAM-pelvis phantom was observed. Rotation of the prostate is often observed in our center as well as in reported literature cases (Deutschmann et al 2011, Huang et al 2015). Huang et al (2015) report a rotation of more than 5° around the left-right axis for 35% of cases. Thus, the overall feasible motion characteristic of the phantom-prostate agrees well with observations in patients in our facility as well as with findings reported in literature. Nevertheless, it would be preferable to be able to reduce the rectal diameter to a minimum and expand it to a volume comparable to the given current maximum to achieve higher variations in rectum wall and prostate displacement.

The different morphological stages of the phantom organs proofed to be well reproducible, ensuring comparability between different experiments. Thus, it can also be assumed that geometry is preserved between measurements in CT and MRI.

4.3. Applicability
The setup of the phantom allows for end-to-end testing of radiotherapy workflows with photons as well as particles including scenarios where treatment adaptation might be necessary due to organ motion. We successfully tested the applicability of the phantom for dosimetric measurements in a simulated treatment of different anatomical states. For this, we used optically stimulated luminescent detectors (OSLDs) in specially created pockets inside the bags on the organ surfaces of bladder and rectum (balloon). Film detectors are being tested for the same application. This way, dosimetry can be performed at the organs at risk in prostate irradiation. For future work, realization of 3D dose measurements of the target is of interest (either for the prostate or for the rectal balloon serving as a tumor model or organs at risk, respectively). This could for example be realized as presented by Gallas et al (2015) using dosimetry gel inside the organ models.

Another application of the phantom is the quality assurance of MR imaging sequences for geometric distortion in a patient-like setup. CT scans can serve as a geometrical ground truth to for example test the spatial accuracy of MR images. Stanescu et al (2012) show that the amplitude and extend of susceptibility induced geometric distortions depends on the shape and size of the enclosed air pockets which is of special importance for the rectum and the adjacent prostate. Wang et al (2013) state in this context that the interplay of organ motion and susceptibility artifacts needs to be investigated for the pelvic region. The presented phantom allows for these demanded investigations of susceptibility induced image artifacts in MRI for different morphological conditions.

The multimodality of the phantom makes it applicable in MR-guided radiotherapy. Kraus et al (2017) show the use of the ADAM-pelvis phantom, as presented in this article, for the validation of synthetic CT creation. MR images of the phantom served as basis for synthetic CTs created by deforming the initial planning CT according to motion vector fields computed with deformable image registration between fractional MR images.

The ADAM-pelvis phantom was furthermore successfully used in investigations in adaptive radiotherapy by Splinter et al (2016). Here, 18 different treatment fractions were simulated with the phantom including six different anatomical states in three different phantom setups created by rotation of the phantom by 0°, 2° and 4°. The fractional CTs were used to compare conventional treatment with treatment adaptation.

Although the phantom is anthropomorphic to a large extent, a phantom can only be a simplification of a patient case. Thus, in the use for testing or developing registration or segmentation algorithms, the results from a phantom study can only give an approximation of the lower limit of uncertainties since they are facing a simplified scenario.

5. Conclusion
The presented ADAM-pelvis phantom describes a solution suitable for investigations in a multimodal (CT, MRI, RT) workflow with anthropomorphic and deformable phantom features. Its applicability in MRgRT has been shown where it adequately mimics a patient in, for example, adaptive treatment scenarios. Additionally, it overcomes limitations present for investigations with patient data. Advantages are given by the use of integrated markers and detectors as well as its reproducibility and controllability in organ deformation. The presented
methodologies might also serve as a basis in the construction of phantoms of other anatomical sites and clinical instances.

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