

Covering Population Variability: Morphing of Computation Anatomical Models

Bryn Lloyd^{1,2(✉)}, Emilio Cherubini³, Silvia Farcito¹, Esra Neufeld^{1,2},
Christian Baumgartner¹, and Niels Kuster^{1,4}

¹ IT'IS Foundation for Research on Information Technologies in Society,
Zürich, Switzerland
lloyd@itis.ethz.ch

² Zurich MedTech, Zürich, Switzerland

³ SPEAG, Zürich, Switzerland

⁴ ETH-Zurich, Zürich, Switzerland
<http://www.itis.ethz.ch>

Abstract. We present a method to change the volume of organs or tissues in computational anatomical models by simulating the human body as a biomechanical solid with initial strains causing local volume shrinkage or expansion. The non-linear hyperelastic material behavior is solved with the finite element method. The bone positions are prescribed and treated as rigid bodies surrounded by elastic soft tissue. A multi-domain mesh defines individual bones and at least one soft tissue region. Each region can have different material properties, volume growth rates or mesh settings. The method can be used to deform complex anatomical models, such as the Virtual Population models. The proposed strategy has been used to parametrize models by different BMI levels, change the volume of selected organs, and modify the posture of anatomical models.

Keywords: Anatomical model · Simulation · Population variability · Obesity · Parametrization · BMI

1 Introduction

Computational anatomical phantoms are increasingly important in academic research and regulatory compliance certification processes. Virtual anatomical models are used to study a variety of scenarios, including, for instance, magnetic resonance imaging (MRI) exposure [9, 17], active and passive implant safety, electromagnetic (EM) field interactions with the peripheral nervous system [18], or passive car safety [24]. Anatomical models have also been used for virtual imaging, for instance to simulate the processes and hardware involved in MRI for designing gradient and radio frequency (RF) coils or pulse sequences [8, 12]. Virtual positron emission tomography (PET) and single-photon emission computed tomography (SPECT) imaging can be performed, e.g., with open-source software and plays a key role in the design of new medical imaging devices, acquisition algorithms, and protocols [20].

The Virtual Population (ViP) [7, 11] is a set of highly detailed computational anatomical models based on MRI data from healthy volunteers segmented at a resolution of $0.5 \times 0.5 \times 0.5$ mm. High quality surface meshes enclosing each of the more than 300 tissues and organs were generated from the segmented label fields [11]. The ViP models already cover important parts of the population variability, including children at different ages, adults, an elderly man, an obese man, and a pregnant woman. However, certain applications require personalized or parametrized models to investigate relationships between differences in morphology or gross anatomical descriptors such as body mass index (BMI), weight, height, or sex. The creation of additional models from new image data could fill some gaps in the population coverage, but would require a significant amount of work to develop. Therefore, strategies are needed to extrapolate from or morph existing models.

Morphing of surface models by means of freeform deformation, e.g., with a control grid, cage-based technique, or thin-plate splines, has been studied in the computer graphics community [19]. Finite element method (FEM) deformable models have been used in virtual reality surgery simulators [23]. Methods to simulate as-rigid-as-possible deformation without a volumetric mesh have been presented in [16, 21] these models are based on assumed homogeneous (tissue) deformation properties with no internal structures such as bones or organs inside the skin surface. Skeleton-based techniques typically allow a human body model to be animated by positioning individual bones that are linked in a hierarchical structure, connected via joints. Bones can be transformed relative to their parents, with their transformations propagated to all children [6], but typically allow only posture to be changed with no alteration to tissue size and shape. Fonseca et al. [10] used MakeHuman [3] and other tools to change the fat volume and posture of a simple human body model consisting of an outer surface and two internal organs for dosimetry evaluations. MakeHuman is open-source software for modeling human characters for computer games and animations [3]. Simple scaling and rigid transformation were used to scale organ surfaces to reference sizes (weights) published by the International Commission on Radiological Protection (ICRP) [4, 15], and similar approaches have been used by others [13]. Ali-Hamadi et al. [1] presented a method to register an anatomical model to a target skin surface while prescribing a fat distribution. The approach is based on registering the surface below the subcutaneous fat between characters using a nonlinear iterative closest point algorithm. The thickness of the subcutaneous fat layer in the target character is assumed to be approximately constant around each bone. While these techniques allow generic deformation of surface models, they deal with homogenous models that are relatively simple compared to the ViP or do not provide strategies for changing, e.g., the volume of specific tissues of anatomical surface meshes in a physiologically realistic way.

In this work, we describe an approach for extending the population coverage of existing models. The method allows an anatomical model to be parameterized, e.g., to high level descriptors such as BMI, weight, or the volume of individual tissues. It is an extension of previous work to change the posture of anatomical

models [6, 14, 22] in which a biomechanical finite element model is used to morph the anatomy on the basis of a set of physically realistic constraints.

2 Methods

We have developed a technique to shrink or expand tissues locally, allowing changes to be made in, e.g., the size of the liver or even the entire subcutaneous fat layer by a specified percentage. The approach treats the body as a deformable hyper-elastic material with rigid bones. Specific tissues can be parametrized by locally prescribing initial strain [22, 25] and thereby controlling the volume. The tissue deformation is constrained by nearby rigid bones and regularized by the surrounding soft elastic tissue. This method can be combined with an approach to change the posture of anatomical models [14]. Bones are moved by prescribing rotations around articulated joints. The bone hierarchy allows relative transformations to be propagated from a parent bone to all children, and the new bone positions are applied as constraints to the biomechanical finite element simulation.

2.1 Hyperelastic Material

A hyperelastic material is defined by its elastic strain energy density W , which is a function of the elastic strain state. It is usually referred to as the energy density and determines the linear or non-linear stress-strain relation and geometric non-linearities. The strain state is often formulated via the right Cauchy-Green deformation tensor \mathbf{C} . For isotropic materials, any state of strain can be described by three independent variables - typically the invariants of the Cauchy-Green tensor.

The strain tensor \mathbf{C} is defined via the deformation gradient \mathbf{F} . In the Lagrangian formulation, the deformation gradient \mathbf{F} can be computed as the displacement vector \mathbf{u} relative to the reference coordinates \mathbf{X}

$$\mathbf{F} = \frac{\partial \mathbf{x}}{\partial \mathbf{X}} = \mathbf{I} + \frac{\partial \mathbf{u}}{\partial \mathbf{X}} \quad (1)$$

where \mathbf{x} is the deformed position, which can be formulated as $\mathbf{x} = \mathbf{X} + \mathbf{u}$. In general the total deformation gradient can be decomposed into elastic and inelastic parts

$$\mathbf{F} = \mathbf{F}_{el} \mathbf{F}_{in} \quad (2)$$

where the inelastic part could be due to initial strain, thermal expansion, or, e.g., plastic deformation. The elastic Cauchy-Green tensor is defined as

$$\mathbf{C}_{el} = \mathbf{F}_{el}^T \mathbf{F}_{el} \quad (3)$$

We use this formulation to introduce a local change in volume, in analogy to thermal expansion, by defining

$$\mathbf{F}_{in} = \mathbf{F}_{vol} = \mathbf{I} \lambda \quad (4)$$

resulting in a volume scale factor of

$$\det(\mathbf{F}_{vol}) = \lambda^3 \quad (5)$$

Accordingly, we can implement different material models with local volume changes by inserting Eqs. 2 and 3 into the strain energy density function, which now depends on an inhomogeneous distribution of $\lambda(\mathbf{X})$, i.e. $W(\lambda(\mathbf{X}))$.

2.2 Numerical Procedure

A variety of hyperelastic material models exist, and each defines a different stress-strain relationship. Currently, we have implemented St. Venant-Kirchhoff, Neo-Hookean, or Mooney-Rivlin material models. We solve for static equilibrium by a non-linear finite element method. The main application of the presented approach is to change the size and shape of individual organs or tissues, e.g., to increase or decrease the amount of subcutaneous adipose tissue (SAT). In this case, we can assume the bones undergo no deformation (the bones are rigid) and define a Dirichlet boundary condition on the surface of the bones. In the simplest case, the displacement of the bones can be set to zero. However, we have developed a more powerful approach, which allows us to move the bones to change the posture of the anatomical model while simultaneously morphing (expanding or shrinking) specific tissue regions.

To solve the deformation on a regular workstation in a reasonable time, the human body model is meshed as coarsely as possible. The use of a tetrahedral mesh with approximately 500K–1.5M elements results in a computation time in the range of 1–5 min. Larger meshes quickly increase the memory consumption and take longer to solve. Obtaining a high quality coarse volumetric mesh is challenging when dealing with complex anatomical models, such as the ViP models [11]. As an example, the SAT surface of the obese model “Fats” contains more than 300K triangles alone, and represents a complex geometry with fine details and thin regions. Directly remeshing the SAT surface frequently introduces self-intersections, which subsequently prevent the tetrahedral mesher from generating a computational mesh. Our current implementation allows us to create a multi-domain (multiple material) tetrahedral mesh by a combination of the following strategies:

- reconstruct tissue surface [5] to remove small features
- simplify the geometry, e.g. replacing a bone surface by a cylinder
- repair self-intersections with heuristics [2] that involve iterative removal of intersecting triangles and closing holes
- locally refine tetrahedral mesh and snapping or smoothing nodes to improve geometric approximation of tissue region.

The displacement field calculated by the FEM on the coarse tetrahedral mesh is interpolated to the vertices of the high resolution surface model. This interpolation or projection step is reasonable, since it can be assumed that the deformation is fairly smooth. The interpolation weights used to interpolate the coarse displacement field on the vertices of the surface mesh are precomputed and stored in the model to further improve the performance of the method.

3 Results

3.1 Parametrization of Subcutaneous Tissue

The morphing method is demonstrated here with the obese ViP model “Fats” (37 years old, 119.5 kg [11]). The body has been meshed with separate SAT, soft (non-SAT) tissue, and rigid bones by means of an adaptive Delaunay mesh generation method with approximately 1.5 million tetrahedral elements. Two different initial strains are simulated, reducing the SAT volume by approximately 60 % ($\lambda = 0.7$) and increasing it by approximately 120 % ($\lambda = 1.3$). Figure 1 depicts “Fats” morphed to different obesity levels.

3.2 Scaling of Organ Sizes

As a second example of the approach, we parametrize an anatomical model by organ volume. Specific organs of the ViP model “Duke” shown in Fig. 2 (34 years old, 70.2 kg) have been morphed to match the values for the adult male referenced in the ICRP Publication 110 [15]. The volume of the heart muscle was scaled by a factor of 1.45, the heart lumen by a factor of 0.5 and the lungs by a factor of 1.27. Figure 3 shows the resulting deformation. Due to the confinement

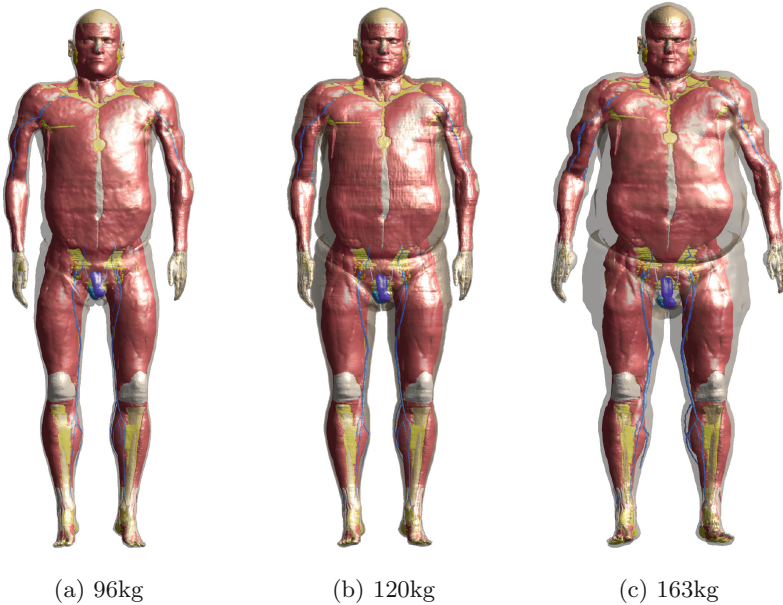


Fig. 1. ViP model “Fats” morphed to different weights and BMI values. The standard “Fats” model is depicted in (b) and weighs approximately 120 kg. The version in Figure (c) was posed slightly to avoid self-intersection of groin and arm regions resulting from the significant increase in adipose tissue. The subcutaneous adipose tissue and skin layer is rendered transparently. The BMI from left to right is 29, 36 and 49

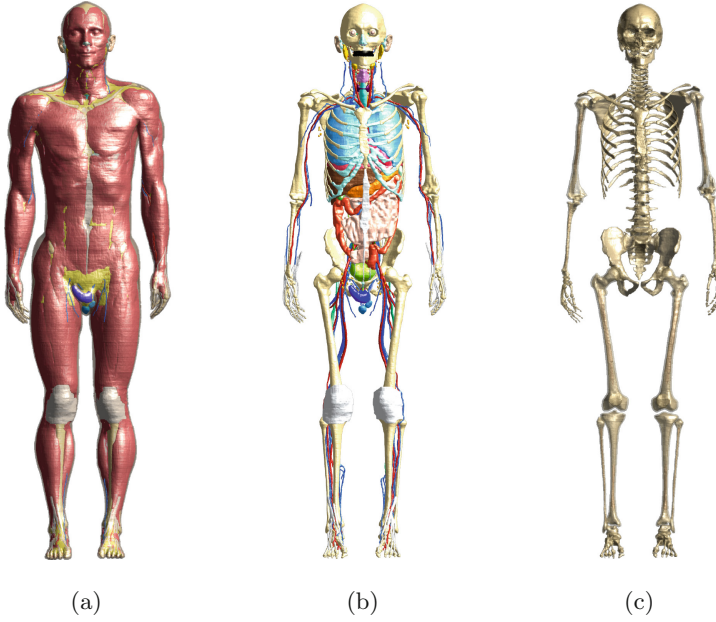


Fig. 2. The ViP model “Duke”

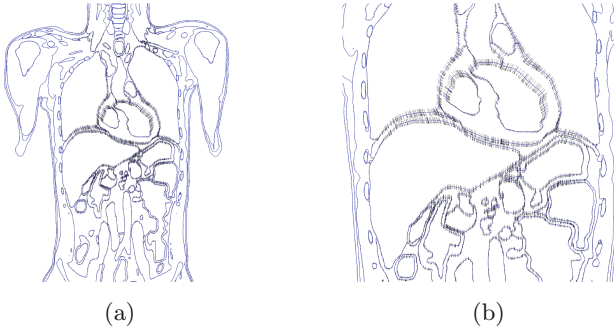


Fig. 3. The ViP model “Duke” with modified organ volumes. The model has been cut to show tissue boundaries with displacement field vectors overlaid. A close-up of the torso section (a) is shown in image (b).

of the rib-cage, the change in volume of the lungs pushes other organs and tissues down. The overall shape of the heart does not change much, because the volume reduction of the lumen is compensated by the increased volume of the heart muscle tissue. The tetrahedral mesh contained approximately 1.2 million elements and is shown in Fig. 4.

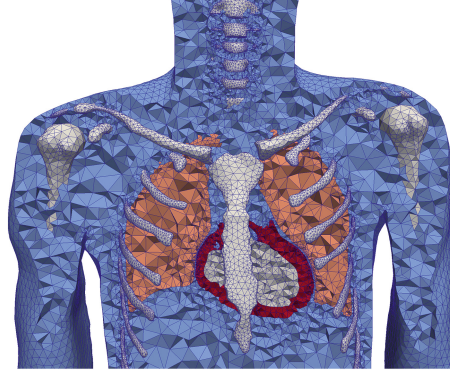


Fig. 4. The adaptive tetrahedral mesh used to morph lung and heart tissues to weights referenced in the ICRP Publication 110 [15].

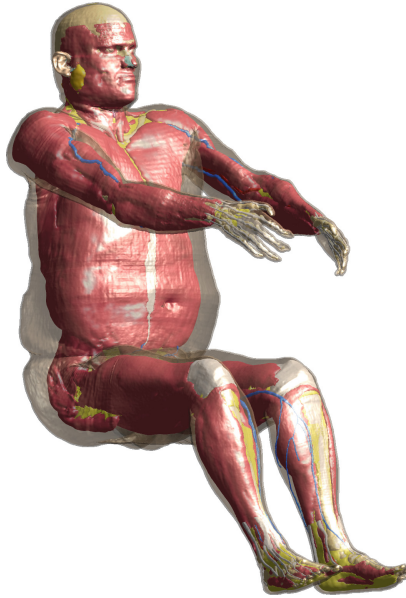


Fig. 5. Image depicting ViP model “Fats” in sitting position. The biomechanical finite element model was used to change the relative position of the bones and simulate the resulting soft tissue deformation. The subcutaneous adipose tissue and skin layer is rendered transparently to show internal structures.

3.3 Posing of Anatomical Model

The biomechanical formulation can also be used to change the posture of the existing model, e.g., for applications in which the safety of humans sitting in a car or at a work place (wireless power transfer, car crash safety, etc.) is investi-

gated. Figure 5 shows “Fats” in a sitting position with arms reaching towards, e.g., a driving wheel. In order to place “Fats” in the driving seat, rotations were prescribed in the hip and knee joint. Similarly the arms and hands/fingers were positioned on the steering wheel by user defined rotations of the humerus and various joints in the hand and fingers. The whole skeleton was posed interactively in a few minutes before the actual deformed model was computed in approximately one minute.

4 Conclusions and Future Work

We have presented a method to parametrize existing detailed anatomical models by treating the human body as being composed of rigid bones and soft elastic tissue, with deformation to balance stresses caused by prescribed spatially varying and tissue-specific initial strains. The method can be used to shrink and expand existing tissues, e.g., fat layers, in a physically realistic way. The various examples described illustrate how the method can be applied to parametrize BMI and the size of individual organs, and how to change the posture of Virtual Population models.

For the BMI parametrization example, we assume isotropic homogenous strains for the SAT, which is clearly a simplification. A detailed literature survey or database of MRI data with good fat contrast is likely to provide more insight into physiologically realistic population distributions of fat tissue, which could be used to define a strain map. The BMI is not only influenced by the SAT distribution but also by the amount of visceral fat. Naturally, the presented method can also scale visceral fat.

A limitation of the current method is that only existing tissue structures can be morphed. For instance, in regions where the adipose tissue is so thin that it was not included in the segmentation, the current approach does not provide a way to create new adipose tissue. Inserting a very thin fat layer between skin and adjacent tissues surfaces might allow us to resolve this issue.

Acknowledgements. The research leading to these results has received funding and support from the Swiss Commission for Technology and Innovation (Project: S4L-CAPITALIS CTI 14930.1 PFLS-LS), COST Action BM1309 and from the European Union’s Seventh Framework Programme (FP7/2007–2013) under grant agreement no. 282891.

References

1. Ali-Hamadi, D., Liu, T., Gilles, B., Kavan, L., Faure, F., Palombi, O., Cani, M.P.: Anatomy transfer. *ACM Trans. Graph. (TOG)* **32**(6), 188 (2013)
2. Attene, M.: A lightweight approach to repairing digitized polygon meshes. *Vis. Comput.* **26**(11), 1393–1406 (2010)
3. Bastioni, M.: The makehuman application open source tool for making 3d characters (2011). www.makehuman.org

4. Broggio, D., Beurrier, J., Bremaud, M., Desbree, A., Farah, J., Huet, C., Franck, D.: Construction of an extended library of adult male 3d models: rationale and results. *Phys. Med. Biol.* **56**(23), 7659 (2011)
5. Calakli, F., Taubin, G.: SSD: smooth signed distance surface reconstruction. In: *Computer Graphics Forum*, vol. 30, pp. 1993–2002. Wiley Online Library (2011)
6. Cherubini, E., Chavannes, N., Kuster, N.: Realistic skeleton based deformation of high-resolution anatomical human models for electromagnetic simulations. In: *The 31st Annual Meeting of the Bioelectromagnetics Society* (2009)
7. Christ, A., Kainz, W., Hahn, E.G., Honegger, K., Zefferer, M., Neufeld, E., Rascher, W., Janka, R., Bautz, W., Chen, J., Kiefer, B., Schmitt, P., Hollenbach, H.P., Shen, J., Oberle, M., Szczerba, D., Kam, A., Guag, J.W., Kuster, N.: The virtual family—development of surface-based anatomical models of two adults and two children for dosimetric simulations. *Phys. Med. Biol.* **55**(2), N23 (2010)
8. Collins, C.M., Wang, Z.: Calculation of radiofrequency electromagnetic fields and their effects in MRI of human subjects. *Magn. Reson. Med.* **65**(5), 1470–1482 (2011)
9. Corcoles, J., Zastrow, E., Kuster, N.: Convex optimization of MRI exposure for mitigation of RF-heating from active medical implants. *Phys. Med. Biol.* **60**(18), 7293 (2015)
10. Fonseca, T.F., Bogaerts, R., Hunt, J., Vanhavere, F.: A methodology to develop computational phantoms with adjustable posture for WBC calibration. *Phys. Med. Biol.* **59**(22), 6811 (2014)
11. Gosselin, M.C., Neufeld, E., Moser, H., Huber, E., Farcito, S., Gerber, L., Jedensjo, M., Hilber, I., Gennaro, F., Lloyd, B.A., Cherubini, E., Szczerba, D., Kainz, W., Kuster, N.: Development of a new generation of high-resolution anatomical models for medical device evaluation: the Virtual Population 3.0. *Phys. Med. Biol.* **59**(18), 5287 (2014)
12. Harris, C.T., Handler, W.B., Chronik, B.A.: Electromagnet design allowing explicit and simultaneous control of minimum wire spacing and field uniformity. *Concepts Magn. Reson. Part B: Magn. Reson. Eng.* **41**(4), 120–129 (2012)
13. Hynčík, L., Nováček, V., Bláha, P., Chvojka, O., Krejčí, P.: On scaling of human body models. *Appl. Comput. Mech.* **1**, 63–76 (2007)
14. Lloyd, B., Cherubini, E., Chavannes, N., Kuster, N.: Realistic physics-based posing of anatomical models for safety evaluations and computational life science in various configurations. In: *BioEM 2016*, June 2016
15. Menzel, H., Clement, C., DeLuca, P.: ICRP publication 110. Realistic reference phantoms: an ICRP/ICRU joint effort. A report of adult reference computational phantoms. *Ann. ICRP* **39**(2), 1 (2009)
16. Müller, M., Heidelberger, B., Teschner, M., Gross, M.: Meshless deformations based on shape matching. *ACM Trans. Graph. (TOG)* **24**, 471–478 (2005)
17. Murbach, M., Neufeld, E., Capstick, M., Kainz, W., Brunner, D.O., Samaras, T., Pruessmann, K.P., Kuster, N.: Thermal tissue damage model analyzed for different whole-body SAR and scan durations for standard MR body coils. *Magn. Reson. Med.* **71**(1), 421–431 (2014)
18. Neufeld, E., Cassara, A., Montanaro, H., Kuster, N., Kainz, W.: Functionalized anatomical models for EM-neuron interaction modeling. *Phys. Med. Biol.*, February 2016
19. Nieto, J.R., Susin, A.: Cage based deformations: a survey. In: Hidalgo, M.G., Torres, A.M., Gmez, J.V. (eds.) *Deformation Models. Lecture Notes in Computational Vision and Biomechanics*, vol. 7, pp. 75–99. Springer, Heidelberg (2013)

20. Santin, G., Staelens, S., Taschereau, R., Descourt, P., Schmidtlein, C., Simon, L., Visvikis, D., Jan, S., Buvat, I.: Evolution of the GATE project: new results and developments. *Nucl. Phys. B Proc. Suppl.* **172**, 101–103 (2007)
21. Sorkine, O., Alexa, M.: As-rigid-as-possible surface modeling. In: *Proceedings of the Fifth Eurographics Symposium on Geometry Processing, SGP 2007*, pp. 109–116. Eurographics Association, Aire-la-Ville (2007)
22. Szczerba, D., Neufeld, E., Zefferer, M., Bhlmann, B., Kuster, N.: FEM based morphing of whole body human models. In: *2011 XXXth URSI of General Assembly and Scientific Symposium*, pp. 1–3. IEEE (2011)
23. Szekely, G., Brechbhlr, C., Dual, J., Enzler, R., Hug, J., Hutter, R., Ironmonger, N., Kauer, M., Meier, V., Niederer, P., Rhomberg, A., Schmid, P., Schweitzer, G., Thaler, M., Vuskovic, V., Tröster, G., Haller, U., Bajka, M.: Virtual reality-based simulation of endoscopic surgery. *Presence: Teleoperators Virtual Environ.* **9**(3), 310–333 (2000)
24. Vezin, P., Verriest, J.P.: Development of a set of numerical human models for safety, June 2005
25. Zienkiewicz, O.C., Taylor, R.L.: *The Finite Element Method for Solid and Structural Mechanics*. Butterworth-Heinemann, August 2005

Simulation and Synthesis in Medical Imaging

First International Workshop, SASHIMI 2016, Held in
Conjunction with MICCAI 2016, Athens, Greece, October
21, 2016, Proceedings

Tsaftaris, S.; Gooya, A.; Frangi, A.F.; Prince, J.L. (Eds.)

2016, X, 178 p. 75 illus., Softcover

ISBN: 978-3-319-46629-3