The Transesophageal Echocardiography Simulator Based on Computed Tomography Images

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Abstract— **Simulators are a new tool in education in many fields, including medicine, where they greatly improve familiarity with medical procedures, reduce costs, and, importantly, cause no harm to patients. This is so in the case of transesophageal echocardiography, in which the use of a simulator facilitates spatial orientation and helps in case studies.**

The aim of the project described in this article is to simulate an examination by transesophageal echocardiography. This research makes use of available computed tomography data to simulate the corresponding echocardiographic view. The article describes the essential characteristics that distinguish these two modalities and the key principles of the wave phenomena that should be considered in the simulation process, taking into account the conditions specific to the echocardiography. The construction of the CT2TEE (Web-based transesophageal echocardiography simulator) is also presented. The considerations include ray-tracing and ray-casting techniques in the context of ultrasound beam and artifact simulation. An important aspect of the interaction with the user is raised.

Index Terms—**biomedical image processing, biomedical informatics, echocardiography, medical simulation**

I. INTRODUCTION

IT is hard to imagine modern medicine without sonography of the heart, described briefly as echocardiography. This technique remains one of the most important tools in cardiology, providing information concerning the heart anatomy and function. There are many other imaging modalities that allow us to visualize the heart, for example, magnetic resonance or computed tomography; however, echocardiography remains in many ways unique because it provides no exposure to radiation, is relatively cheap, and, in the case of transthoracic echocardiography, has no known side effects.

The equipment set used for ultrasonography (US) consists of:

- a sonographic probe containing a transducer, which is able to generate and detect ultrasound waves, and
- a central unit equipped with a monitor and a set of switches for changing the transducer parameters and display options.

Depending on the placement of the sonographic probe during the examination, there are several variants of echocardiography [1], [2]:

- transthoracic: the transducer is placed on the surface of the chest wall
- transesophageal: the transducer is placed in the esophagus
- intravascular, intracardiac, and epicardiac: the transducer is placed, respectively, in the large vessels, in the heart itself, and on the heart surface. These examinations are performed rarely in comparison to the transthoracic and the transesophageal approaches.

In this paper, we concentrate on transesophageal echocardiography (TEE).

Independent of the placement of the echocardiographic probe, different visualization techniques can be used; for example, one-, two-, or three-dimensional echocardiography (M-Mode, 2D, 3D) and techniques based on the Doppler principle. Further in this paper, we will discuss the 2D technique, which is the integral visualization mode in every echocardiographic examination. In the case of 2D echocardiography, the transducer sends a set of ultrasound waves in several directions in one plane and then detects the reflected waves. The ultrasound waves are reflected on the outer edge of tissues having different sound impedances. The speed of sound in the soft tissues is similar; therefore, a constant speed of sound is assumed by any US scanner system. Knowing the time elapsed between the sending of the signal and the detection of the reflected wave, the distance to the point where the reflection occurred can be calculated. Most of the tissues have distinct properties in reflecting the ultrasound waves; therefore, echocardiography allows not only detecting the outer edges between the different tissues but also distinguishing between tissues, such connective tissue, muscle tissue, and fat tissue. Using this very briefly described principle in 2D echocardiography, we are able to get a

reconstruction of a cross section of the heart in a given plane.

The examination itself can be conducted by a physician or an echocardiographic technician, both of whom have to undergo a long period of training before being able to independently perform the echocardiographic examination and then interpret the results [3].

There are no known side effects of transthoracic echocardiography. The procedure is not unpleasant for the patient so there is no time pressure on the trainee in conducting the examination, and in case of insufficient image quality, an experienced echocardiographer can repeat the examination.

This is not the case with transesophageal echocardiography, where the patient has to ingest a probe wrapped in a cable of approximately 1cm diameter (Fig. 1); at the end of the probe is a sonographic transducer. To obtain all the necessary views of the heart, the probe has to be moved (backward or forward) and turned around. The tip of the probe can also be bent with the use of knobs located on the handle of the probe [4].
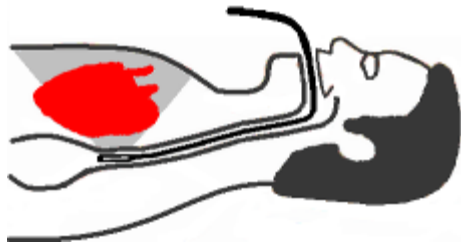

Fig. 1. Position of the probe during TEE examination.

Additionally, the transducer can be rotated inside the probe to increase the number of possible views. These features will be discussed in detail later in this article.

The ingestion and manipulations of the probe are usually highly unpleasant for the patient. The examination may cause small injuries to the throat, resulting in infection and other complications during and after the examination. Because of this, transesophageal echocardiography should be performed quickly but very comprehensibly, and it cannot be performed for training purposes only. All these factors make transesophageal echocardiography much more difficult to learn. Thus, the use of TEE simulators that could speed up and complete the learning process seems very promising.

A PubMed search revealed two working groups that have developed a TEE simulator [5]-[8]. Both describe a mannequin-based simulator, which is used in a limited number of medical training centers. Thus, although a high number of people have to be trained every year, only a few will be exposed to training on the use of a TEE simulator.

We therefore developed the CT2TEE, a novel, Internet-based TEE simulator based on computed tomography (CT) images [9], which will be discussed further in this paper.

Another Web-based project for training on the use of TEE is the virtual TEE simulator of the Perioperative Interactive Education Group, Department of Anesthesia and Pain Management, Toronto General Hospital, (http://pie.med.utoronto.ca/tee/index.htm). The main advantage of this platform is the demonstration of a 3D heart model in relation to the TEE probe. The user can choose from a limited number (20) of predefined views. Therefore, this platform has the features of an online teaching tool rather than of an online simulator.

A general ultrasound simulator, Field II, (http://server.oersted.dtu.dk/personal/jaj/field/), offers an excellent platform for studying physical tissue properties and the simulation of acoustic effects. Unfortunately, because of the accuracy of the simulation, it performs extensive computations and requires up to several minutes to generate one view, making it unsuitable for real-time simulation.

In recent years, there have been other approaches to simulating ultrasound, especially using parallel high-performance environments of graphical processing units (GPU) [10]-[14]. The method presented in this paper features some simplification and does not require high computational power.

Comparisons of simulation methods are presented in [15]-[17], in which the authors discussed different simulators, including the CT2TEE.

II. METHODS AND DATA

The construction of a TEE simulator requires several steps, as discussed in the following sections.

A. Input data

Several sources of data can be used to build a 3D heart model. In his models, Michael Weidenbach used data collected during 3D echocardiography [5]-[7]. However, this method has several limitations. First, the resolution of such 3D data is markedly lower than that in 2D echocardiography. This is why the 2D images reconstructed from these data sets are blurred. There are also many heart structures that, for other reasons, will not be seen at all in such reconstructions [18].

The transesophageal echocardiography simulator described by Bose et al. [8] uses a different principle; that is, an artificial model of an anatomically correctly beating heart was built using software. This concept offers a significantly higher spatial resolution. The trainee can, however, work with only one heart model and will not be able to appreciate the anatomical differences in patients or train in other cases (for example, congenital heart diseases).

Therefore, we decided to use data from a computed tomography (CT) scan of the heart. A similar solution was recently presented by Kutter et al. in [10], where the authors described a GPU-based method for the simulation of ultrasound images from volumetric CT. They created a ray-based model of the ultrasound to generate view-dependent ultrasonic effects, such as occlusions, large-scale reflections, and attenuation, combined with speckle patterns derived from preprocessing the CT image using a wave-based model of ultrasound propagation in soft tissue. The main assumptions of CT-to-US transformations are presented in [19].

There are two ways of using a computed tomography scan of the heart to create a heart model:

- ECG (electrocardiography)-triggered CT and
- multiphase contrast-enhanced helical CT.

The first method, which is older, allows creating a static 3D model of the heart that can be used in preliminary exercises with the simulator.

The second and newer method allows creating a dynamic 4D heart model. The motion can be achieved by analyzing subsequent CT blocks for each acquired heart phase. This kind of CT equipment is also ECG-triggered and allows scanning each phase of the cardiac cycle with a 5% accuracy rate. The output data are subsequent CT 3D blocks that can be used to generate successive frames of the simulated US.

CT tests were performed on patients due to clinical reasons and not for the purpose of the present study. We used data sets taken from studies performed for the assessment of coronary arteries, which therefore offer a high spatial resolution and high contrast differences between blood and soft tissue. Preliminary tests were performed using Siemens SOMATOM Definition Flash. We used standard coronary CT protocols with a reported radiation exposure of 3-20 mSv, a standard tube voltage of 120 kVp, and a voxel size of 0.4x0.4x0.4mm [20].

B. 3D model of the heart

Because of the unlimited number of cross-section planes in the TEE simulation, creating prepared views for each angle is unproductive and even unfeasible. Therefore, a 3D heart model first has to be created to simulate 2D US views.

At every step of the image acquisition, a 2D CT image in the plane of the CT scanner is created (Fig. 2). To reduce the motion artifacts, the examination is performed under breath-hold. The capturing of images is ECG-gated, ensuring that different images are recorded in similar phases of the heart cycle. After an image has been acquired in one plane, the couch is moved slightly (ca 1mm in the z-axis), and the next
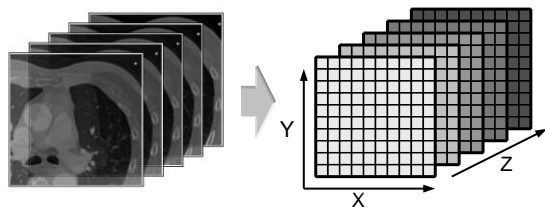

Fig. 2. A set of CT data. Each next slide of the CT set is taken as the subsequent surface for the Z value. The values (X, Y) denote the pixels of the surface.

image is taken. Figure 2 shows a block or set of CT data acquired in the manner used to construct the 3D model in our application.

The multiphase contrast-enhanced helical CT creates volumetric CT data in a similar way. The gantry rotates very quickly around the patient while the table/patient moves through the gantry in a continuous motion. The result is a series of 3D CT blocks that are images of the heart in subsequent moments of time. Because of the harmful radiation, the examination is limited to a single cardiac cycle.

The DICOM standard was chosen as the input format. It allows loading the data generated by various types of CT equipment. The resolution of CT images is usually 12 bits per pixel. The block of CT data is kept at the 12-bit resolution,

which gives high-precision calculations.

To create an appropriate 3D model, a 3D mesh can be extracted from the CT images, which increases the accuracy of computations. The resolutions of CT and ultrasound images are comparable; therefore, it is satisfactory to use raster data with trilinear interpolation instead of a 3D mesh.

Another issue is the relationship between the dimensions of the voxels and those of the model. CT scans allow us to take shots throughout the whole chest or of the volume containing the heart only. Therefore, the X and Y dimensions must be scaled proportionally to the target size of the model of the heart. Similarly, the distance between CT slices should be included in the z-axis scale.

C. Virtual probe trajectory

In an actual TEE examination, the probe is placed in the esophagus and can be moved along it. In the TEE simulation, the trajectory of the probe movement should be set on the CT images by an expert physician (e.g., a radiologist). This should be done only once, at the stage of the creation of the simulation set. Thus, the trajectory of the virtual TEE probe along the specified line connecting the esophagus enables changing the position of the virtual probe within the range of the CT data used.

D. 4D model of the heart

To provide motion in the echocardiography simulation, multiphase CT data can be used. The procedure for the simulation of the 4D model is analogous to that of the 3D model. There are a selected number of 3D CT data blocks that correspond to the heart cycle phases. An animation frame is created in the process of the simulation using an adequate 3D CT data block. Switching the frames of heart cycles provides motion in the simulator. The principles of simulation for a single animation frame are the same as those for a static 3D model.

E. Reconstruction of 2D slices for US

The simulation process of 2D TEE begins with the creation of a view plane for a desired projection. For this purpose, the position of the probe head and the plane of projection should be determined. The head position is based on the depth of probe insertion (p). This value and virtual esophagus data are used to set the coordinates for a 3D model (x, y, z). To define the projection plane, three characteristic angles of the transmitter should be included:

- α (X, Y) – rotation (moving the TEE probe around its long axis, range of 0:360 degrees)
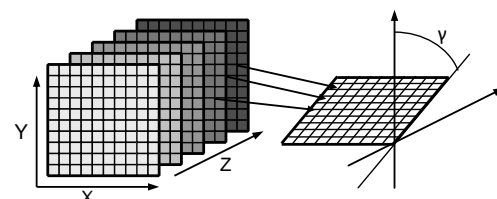

Fig. 3. Example of a 2D reconstruction from a 3D data set by changing the γ – angle value.

- β (Y,Z) – deflection (bending the end of the TEE probe anteriorly or posteriorly, range of -30:30 degrees)
- γ (X,Z) – plane angulation (axial rotation of the transducer at the tip of the TEE probe, range of 0:180 degrees; Fig. 3)

The last parameter is scan depth (d), which is related to the desired magnification of the view.

All these variables (p, α, β, γ, d) create a space of control parameters for the virtual TEE probe. The user should be able to set these parameters on the probe in an interactive process. Therefore, an adequate graphical interface is proposed (Fig. 13).

III. ULTRASOUND SIMULATION USING CT

To construct a simulator that uses the CT scan, the comparison of these different modalities in terms of image creation is needed.

A. Physical basis for the ultrasound simulation

Based on the projection of the CT data for the given angles, an ultrasound projection can be simulated. This requires consideration of the following wave phenomena:

- reflection,
- attenuation,
- refraction, and
- diffraction.

Reflection

The reflection of ultrasonic waves can be considered in two categories:

- specular – reflection from large objects at the surface of two tissues of different acoustic impedances (z_1, z_2); the amount of the reflected wave energy R can be determined by the Fresnel formula (1):

$$R = \frac{z_1 - z_2}{z_1 + z_2} \qquad (1)$$

In case of large differences in acoustic impedance between media M1 and M2, a wave can be reflected completely (e.g., a tissue and a bone, tissue and air in the lungs).

- scattered – reflection from small objects (i.e., blood cells); this shows the shape/pattern of a tissue and can be a source of noise.

The values of the acoustic impedances (z, reduced to r_a) of soft tissues and blood are similar (see formula (2) and Table I). This is very helpful in simulations in the cardiologic domain.

$$r_a = \rho * V \qquad (2)$$

Attenuation

The attenuation of ultrasound waves in tissue consists of the phenomena of reflection (described above), absorption, and scattering. The signal of the beam is increasingly attenuated with the progress of the reflection time. Although the ultrasound equipment uses the time gain compensation to counteract this phenomenon, there are some noticeable artifacts when the beam passes through muscles of varying thicknesses. The attenuation in blood and soft tissue differs (Table I), and this should be taken into account during the construction of the simulator.

Refraction

In cases when media of different velocities are encountered, part of the beam energy is not reflected but is transmitted and undergoes refraction or bending. The beam angle can change according to Snell's law:

$$\frac{\sin \theta_1}{\sin \theta_2} = \frac{c_1}{c_2} \qquad (3)$$

The change in angle (θ_2, θ_1) is correlated with the ratio of the sound velocities in these media (c_1, c_2).

The sound velocities in blood and soft tissue are very close to each other, and the refraction phenomenon in this case is insignificant. This fact is essential in diagnosis, whereas the accuracy of simulation is less important. Therefore, in the simulation of echocardiography, the refraction phenomenon can be omitted. Otherwise, complex ray tracing should be done because the waves can go beyond the plane of projection.

Diffraction

Diffraction involves a change in the direction of the ultrasound waves as they pass around small objects in their path. This causes acoustic shadowing. It is an important phenomenon in cardiac diagnosis, allowing the detection of calcifications in the heart.

B. Generating the TEE view from CT projection

The mechanism of generation of TEE images using CT projection consists of the following stages:

- brightness level transformation and
- simulation of the ultrasound beam.

Transformation function for brightness levels

Both CT and TEE deal with several types of tissues and structures, most of which have different brightness levels in the

TABLE I
PHYSICAL PROPERTIES OF SELECTED TISSUES, [21], [22]

Tissue type	density ρ – [kg/m3]	velocity V – [m/s]	Attenuation dB/[cm*MHz]
Soft tissue	1040±100	1540±15	0.5±0.05
Blood	1050±100	1570±30	<0.1

TABLE II
BRIGHTNESS COMPARISON OF DIFFERENT TYPES OF TISSUE IN CT AND TEE

Tissue type	CT brightness level	TEE brightness level
blood (with radiologic contrast agent)	very high	low
connective tissue	moderate	high
heart muscle	moderate	moderate
lungs, airways	very low	not visible (noise)
bones	very high	not visible (noise)

two imaging modalities (Table II).

CT images acquired from different patients differ in terms of the brightness bands for the types of tissues listed in Table II. Additionally, the usually applied radiologic contrast agent causes an overlap of brightness levels for several tissue types, such as bones and blood with contrast agent. Another problem during the examination is the changing contrast concentration over time. Therefore, it is difficult to create a fully automated algorithm that can clearly mark the boundaries between heart tissues and to perform a fully automated conversion of brightness bands to TEE-like images for CT data from different examinations, different patients, and various equipment vendors. In these cases, the histograms differ significantly (Fig. 4). The histograms are also influenced by
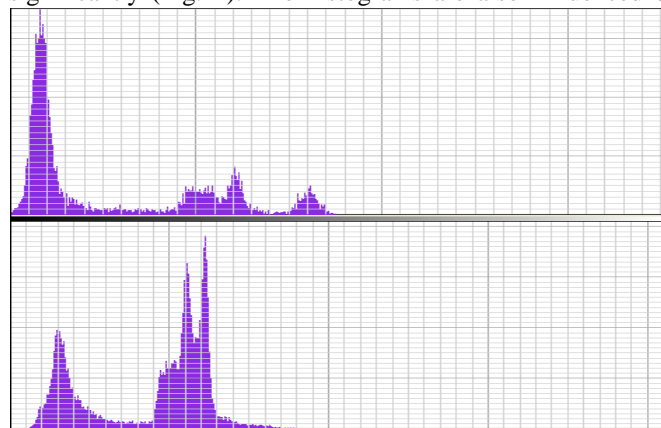
Fig. 4. Examples of histograms for a projection for CT examinations performed using two different types of equipment (different from that in Fig. 5).

the location and magnification of the images (depending on the content of the image).

Therefore, using manual adjustment of the brightness levels, a nonlinear transformation function is performed in the preparation phase for every CT data set. That is, the physician who prepares the set must manually adjust the brightness ranges for selected tissues in both modalities. With the use of a grayscale histogram of images, the parameters of the
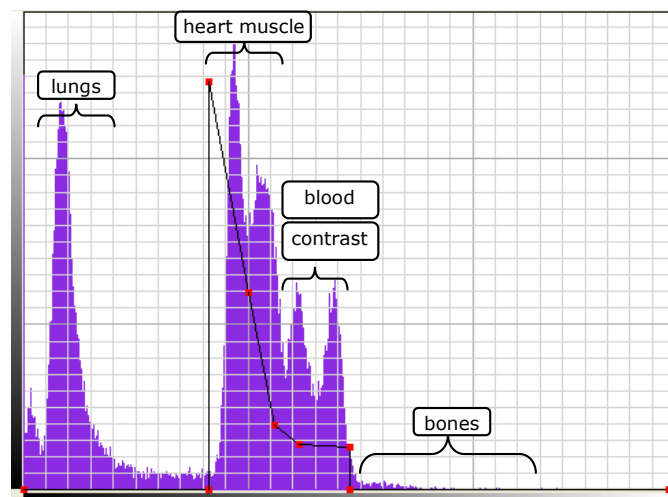

Fig. 5. An example of a grayscale histogram of a CT image slice (normalized to 100%) with plotted transformation function (thin black line with red points). The horizontal axis represents the input brightness level, and the vertical axis the output brightness level.

transformation function are adjusted to optimize the TEE (Fig. 5). Although the brightness ranges of the tissue types in the two modalities are different, the shapes and structures (patterns) of the tissues remain the same. Therefore, the construction of the transformation function can be limited to compare the ranges for the tissues only.

All CT studies were performed using a contrast agent.
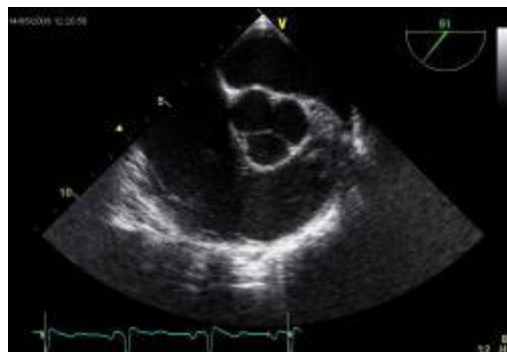

Fig. 6. An actual real TEE view with the aortic valve in the central position (Mercedes-Benz sign).
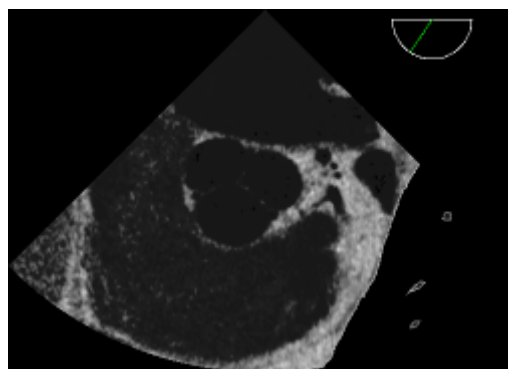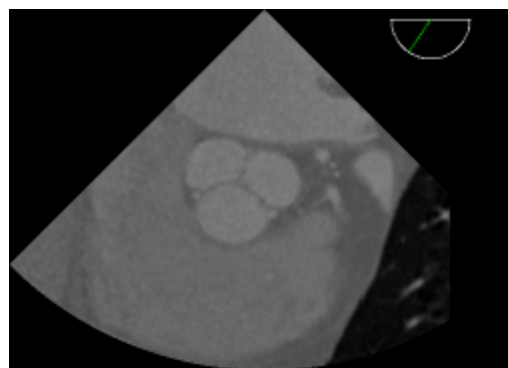

Fig. 7. An example of an input computed tomography image (top) and of an output of the transformation function for TEE (bottom); the brightness of tissues is based on the presented rules and transformation (Fig. 5).

Several data sets have been tried, but only those performed with contrast offered sufficient visualization quality.

Figure 6 shows an example of an actual projection, whereas Fig. 7 presents an example of a corresponding CT projection and transformation into a TEE-like view.

Every CT data set has to be prepared by a cardiologist to achieve the optimal transformation settings (Fig. 5). However, after such adjustment, several structures, such as bones or air-containing tissues, are still visible, thus making the additional steps described below necessary.

C. Simulation of the ultrasound beam in the CT2TEE

The simulation of the ultrasound beam is a very important part of the project and corresponds to the wave phenomena (reflection, attenuation, refraction, and diffraction). To address this issue, a ray-casting or ray-tracing algorithm should be used.

The ray-casting algorithm is simple to implement and not time-consuming. It is based on the tracing of each ray along the direction that the ray follows. Ray casting enables the simulation of attenuation (according to the values in Table I) and reflections.

The ray-tracing algorithm is more complex to implement and can be much more time-consuming [23]. It traces all rays, taking into account the wave phenomena (reflection, attenuation, refraction, and diffraction). One study [24] showed that implementing all phenomena in a component environment takes so much time that it does not permit real-time simulation.

In the case of cardiac simulation, some features are insignificant (described in the previous section). Therefore, the ray-casting technique can be regarded as more competent.

D. Simulation of the ultrasound artifacts in the CT2TEE

In the CT2TEE, we decided to address the major sonographic artifacts, such as US shadow, attenuation errors, US-like noise, and simple US wave diffraction. The artifacts associated with velocity errors are not taken into account because of the similar sound velocities in blood and muscles. Other artifacts (reverberations) are not frequent in transesophageal echocardiography, and their absence in the simulator does not decrease the educational value of the method.

US shadow and wave refraction

The CT2TEE application addresses the ultrasound shadow by a simple function based on the ray-casting algorithm. For every point along each radius, starting with the esophagus position in the current projection, the condition of casting is checked. If the point is an obstacle (bone, air), it is marked as a black hole (value of 0) according to the transformation function. The values for the following points in this radius are modeled graphically by the function presented in Fig. 8. These
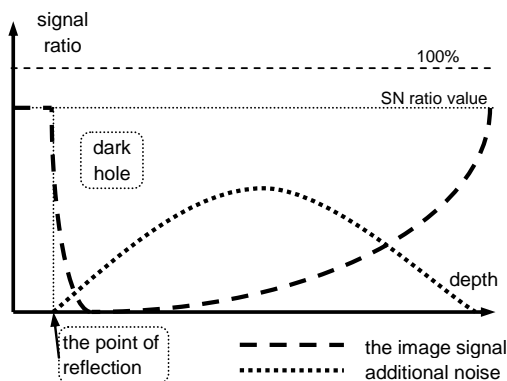
points are seen on the image reconstruction as dark holes after bone or air. The length of the applied function is proportional to the length of the obstacle. Because the obstacles (mostly calcifications) are oval-shaped rather than long and thin, a
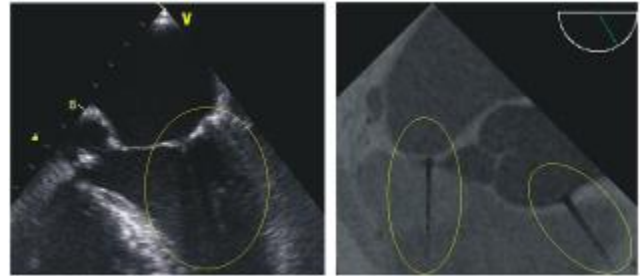

Fig. 9. An example of a US shadow caused by small calcifications on the mitral valve (left), and a simulation of US shadows in the CT2TEE application (right).

comet-like shadow is usually created (Fig. 9).

US attenuation

The ultrasound beam is attenuated as it propagates through tissue. The attenuation is several times greater for heart muscles than for blood (Table I). This causes diminished brightness for rays that have passed a longer distance through muscles. Therefore, the simulated sound beam power is decreased by the integral of the attenuation in the path of the ray for blood and muscle tissues, respectively, as shown in Table I.

US-like noise

Ultrasound images have a characteristic, concentric noise. To mimic it in real time, the CT2TEE application creates a rectangular table with generated white noise and overlaps it onto the current view by transformation from a Cartesian coordinate system to a polar coordinate system (Fig. 10). For
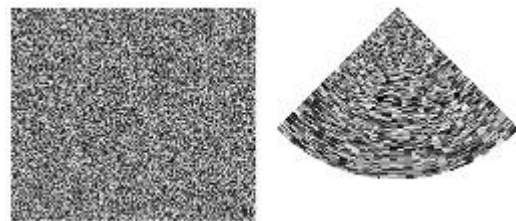

Fig. 10. Noise transformation from a Cartesian coordinate system to a polar coordinate system.

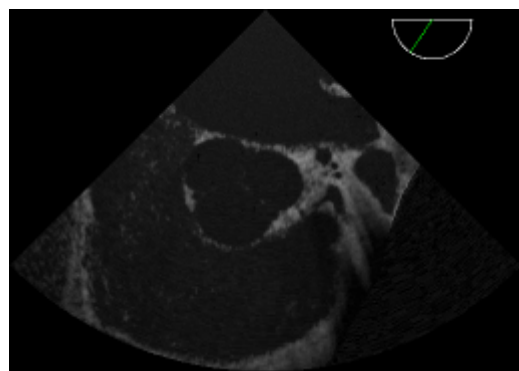Fig. 8. Graphical presentation of the function simulating US shadow

Fig. 11. The output of the US simulation for the projection presented in Fig. 7
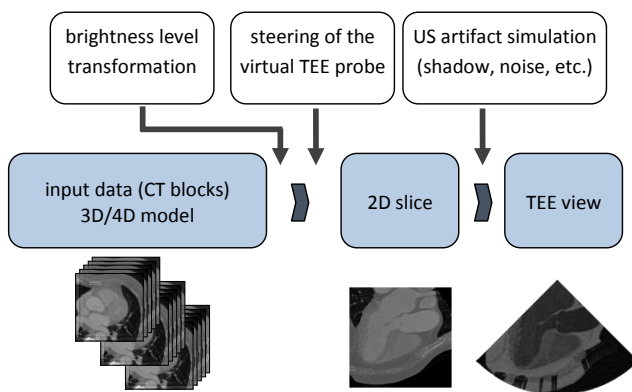
Fig. 12. A diagram of the data processing in the CT2TEE application.

each projection, a seed of a random generator is calculated to keep sight of reality. The value of the noise is added to each point of the simulation view according to the level of signal-to-noise ratio defined by the operator who prepared the CT set.

E. Simulation summary

Figure 11 shows the final output using the ultrasound beam simulation for the projection presented in Fig. 7.

Figure 12 presents a general diagram of the simulation calculations. It should be noted that due to time optimization reasons, the transformation of the brightness levels can be performed only once, at the beginning, for the entire model of the heart. Motion is achieved in the simulation by generating frames for the subsequent phases of the cardiac cycle. Despite its high memory consumption, this method is very effective.

IV. THE STRUCTURE OF THE CT2TEE APPLICATION

The CT2TEE application was created in the Microsoft .NET 2.0 component environment. This framework allows the creation of a standalone (thick client) program and the parallel development of a Web version (thin client) in a fairly straightforward way [25]. However, the standard Win32 application has much more functionality and is faster than the online version.

The standalone version allows creating sets for the TEE simulation and performing real-time simulation. Generating a single frame on a single processor takes under 100ms on an average PC or notebook. The computations are well parallelized, e.g., using CPU 4 cores takes 25ms to generate a single frame. This is a very good result for the calculation using the component platforms [24].

The online version of the CT2TEE was created as an ASP.NET service that returns a simulated projection for the requested parameters of the virtual TEE probe. These settings correspond to variables ($p, \alpha, \beta, \gamma, d$) that create a space of control parameters for the TEE probe. The control panel, written in JavaScript, allows the user to set the probe position and angles (Fig. 13). Each change in the virtual probe position on the client side initiates a request to the server for an image that is generated based on the given projection parameters. Two output formats are available: a JPG (smaller file size) and a GIF (animated) image.

The online CT2TEE simulator is available at the following Internet address:

http://www.ct2tee.agh.edu.pl

V. CONCLUSIONS AND FUTURE WORK

Since July 2009, nearly 13,000 users have visited the CT2TEE home page, and more than 430,000 projections have been generated. The first stage of development of the CT2TEE has already been completed. The further development of the project aims to provide a transthoracic echocardiography (TTE) simulator. In our opinion, however, even at the current stage of development, both versions of the CT2TEE simulator are valuable teaching tools that allow the simplification and shortening of the complex process of learning TEE.

Transesophageal echocardiography is a manual task, and the steering of the probe may be a major obstacle for beginners. Professional TEE simulators that make use of a mannequin may therefore have additional educational value compared to our simulator. Nevertheless, we believe that our simulator may help users to understand the rules on generating appropriate TEE views and may facilitate the process of TEE learning. It does not, however, serve as a substitute for hands-on training.

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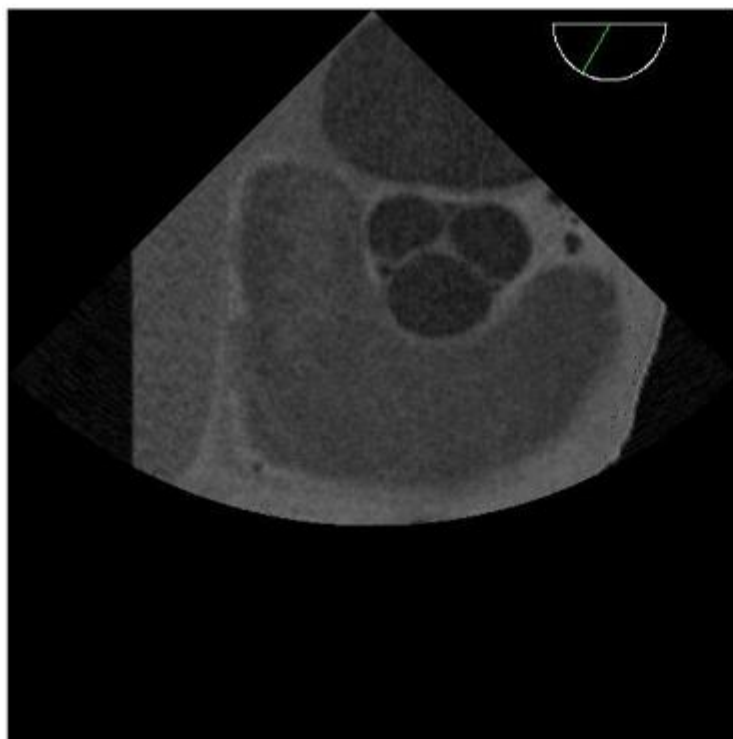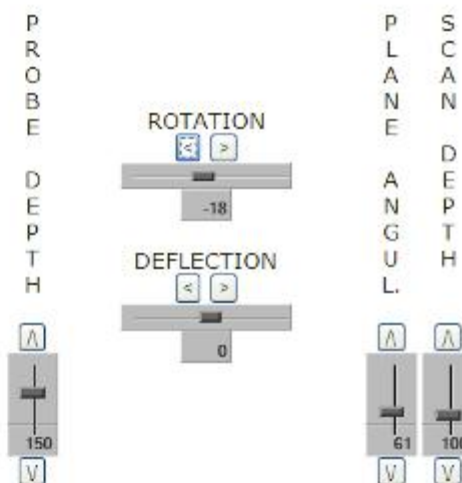
Fig. 13. The control panel in the online version of the CT2TEE