Basic principles and practical application

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Current real-time 3D echocardiography (RT3DE) evolved from original volumetric three-dimensional (3D) echocardiography. Original volumetric 3D echocardiography described the acquisition of 3D datasets in real time at a rate of about 15–20 3D datasets per second using sparse matrix array transducer technology [1][2][3][4], thus, eliminating time-consuming 3D reconstruction of two-dimensional (2D) image planes which were prone to interpolation artifacts (▶ Chapter 1) [5][6][7][8]. Such pyramid-shaped, volumetric 3D datasets, once acquired, could be further analyzed by cropping and slicing.

Recent developments in RT3DE, however, go beyond this by allowing 3D visualization of volume rendered images during live scanning, commonly called live 3D echocardiography [9]. Thus, current RT3DE can be used as an integral part of routine echocardiographic examinations allowing rapid change between 2D cross-sectional views and 3D visualization with the press of a button. Because RT3DE works differently in many regards compared to conventional 2D imaging, the following chapter provides a comprehensive overview of the fundamentals of current real-time 3D technology, including the basic principles of volume data acquisition and requirements of 3D probes, followed by a practical introduction to 3D data acquisition and analysis. The wide spectrum of clinical indications for 3D examination will be covered in the subsequent chapters on special applications and clinical questions.

Today, RT3DE can be used as an integral part of routine echocardiographic examinations.

2.1 How real-time 3D ultrasound works

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Three-dimensional ultrasound can be challenging for clinicians to interpret. It is also challenging for ultrasound engineers to design. No other ultrasound mode has more limitations and more need for improved frame rates, resolution, image quality, and volume size. These constraints impact echocardiography the most, because of the high frame rates required to capture the motion of the heart. The intention of this section is to provide insights into the constraints and trade-offs faced by ultrasound engineers developing 3D ultrasound modes and some of the methods used to address these challenges.

2.1.1 It's all about the transmit beams

Whether 2D, 3D, color Doppler, or B-mode, the basic unit for all ultrasound datasets is the ultrasound beam. When envisioning this ultrasound beam (particularly on transmit), a good analogy would be the light beam from a flashlight (Fig. 2.1).

Scanning a room with a flashlight beam is a good analogy to 2D and 3D ultrasound.

Imagine standing in a dark room with nothing but a flashlight with a relatively narrow beam. To see what is on a shelf on the other side of the room, one would simply point the flashlight beam on the left side of the room and scan the shelf from left to right. The light is emitted from the flashlight, transmitted across the room, reflected off the shelf, and finally received by the eye (Fig. 2.2). This is a good analogy to 2D ultrasound.

However, the room is three-dimensional. Perhaps there is a table and a chair and more bookshelves in the room. For the original scan of the bookshelf, the narrow light beam had a limited diameter, say 10 cm, so all of these other structures were missed.

One approach to scanning the entire room between the flashlight and the wall is to initially place the beam in the upper left corner of the wall. Then traverse the beam from left to right (just like in a 2D scan).

When the beam reaches the upper right edge of the wall, start on the left side again, but with the beam 10 cm lower than the first pass. Repeat this sequence until the entire wall and other structures are scanned from ceiling to floor (Fig. 2.3). Such a sequence is nearly identical to the scan sequence used in 3D ultrasound equipment (Fig. 2.4) where each scan line corresponds to a different direction in space of the beam emitted from the flashlight.

By integrating the light over the entire scan, the entire shelf and room can be evaluated (Fig. 2.5). On 3D ultrasound equipment, this integration occurs by digitizing the returning sound echoes and storing them in the system's memory. Thus, a room that is 5 m wide by 3 m tall and a light beam that is 10 cm in diameter would require 50 light beams across and 30 light beams high to scan the entire wall. Note that such numbers are not atypical of the number of transmit beams used in both 2D and 3D echocardiography. Using typical echocardiography times, it takes about 0.2 ms for the sound to emanate from the face of the probe, hit a target 15 cm deep, and return to the face of the probe. Therefore, 50 transmit beam emissions require 10 ms to scan the wall, resulting in a sustained frame rate of up to 100 Hz. This is just for the 2D B-mode.

To preserve the same image quality and line density used in 2D, 3D would require 50 horizontal/lateral beams by 30 vertical/elevation beams or 1500 transmit beams per volume, with a total scan time of 300 ms. The resulting low 3.3 Hz frame rate would be unacceptable.

Alternatively, given that adult echocardiography typically »requires« frame rates in excess of 20 Hz (50 ms), how many transmit

Fig. 2.1 Flashlight beam: analogy for an acoustic transmit beam

Fig. 2.2 Scan of a bookshelf is analogous to a 2D B-mode scan. The illustration shows a room with bookshelf and a light beam emitted from a flashlight, transmitted through the room, and reflected off the shelf. By traversing the beam from the left to the right (*red arrow*) allows the entire upper bookshelf (just like in a 2D scan) to be scanned

Fig. 2.3 The 3D scan of bookshelf illustrating 3D ultrasound scan. The 3D scan of the room does not only illuminate the bookshelf but all objects in the three-dimensional room (like the small tree)

events (beams) could be supported in a live volume dataset? The typical transthoracic echo depth of 15 cm (with a 0.2 ms echo transit time) would limit the volume to only 250 transmit events. An example scan grid of 25 transmit beams in the lateral dimension (the 2D scan plane) by 10 transmit beams in the elevation dimension would severely sacrifice the image quality as compared to 2D with a similar lateral span. This is because the 3D line spacing would have to be significantly coarser.

This is essentially the 3D problem. One can either sacrifice frame rate for image quality (resolution) or resolution for frame rate. It is very difficult to have both. The options and trade-offs to address this problem are as follows:

- Limit the total number of beams by scanning a smaller volume (3D zoom mode).
- Tolerate the degradation in frame rate.
- Use the patient's ECG signal to acquire smaller »subvolumes« for each R-to-R interval, and then concatenate (stitch) these subvolumes to produce the larger volume (also termed full volume dataset).
- Employ spatial aliasing. This is where the ultrasound beams (on both transmit and receive) are spread far enough apart that there are gaps between the beams that are not adequately sampled. In the bookshelf/flashlight analogy, this would be illustrated by moving the 10 cm diameter beam 20 cm in each interrogation. Clearly the books located between beams would not be well illuminated. In echocardiography, this will either be observed as a scintillation of the speckle pattern (observed on older 2D B-mode scan-

Fig. 2.4 Transmit scan lines emanating from a matrix transesophageal transducer

ners) or a blurring of small structures. This blurring effect can often be exacerbated by selecting the low line density option available on most 3D ultrasound machines.

- Broaden the transmit beam. In this case, the intent would be to increase the diameter of the beam from 10 cm to 20 cm. This would allow one to scan the wall twice as fast in both directions (4x faster for a volume scan). But as will be discussed later, such a technique is also not without compromise.

All of these »solutions« are employed to varying degrees by all 3D ultrasound manufacturers.

2.1.2 The need for bigger, better, faster

Anyone who has ever used 3D echocardiography clinically will inevitably request larger volumes (bigger), improved image quality (better), and faster frame rates (faster).

The 3D volumes in echocardiography are often degraded due to sparse, poorly focused, low intensity transmit beams. In the attempt to maintain acceptable frame rates, the ultrasound engineer will both spread and broaden the transmit beams. To broaden the transmit beam, there are really only two options: either reduce the active aperture (transmit on fewer elements) or defocus the beam (e.g., focus at infinity).

Perhaps worse is the impact on sensitivity and penetration. By broadening the beam (Fig. 2.6), the transmitted ultrasound energy is spread over a larger area. However, this is particularly constraining for tissue harmonic imaging. In order for an adequate harmonic wave to be generated, the intensity needs to be above a certain threshold. Transesophageal images, which predominantly rely on nonharmonic imaging modes, are less susceptible to the sensitivity issues caused by transmit beam broadening. Therefore, in order to maintain adequate sensitivity and image quality when scanning fast (coarse transmit beam spacing), the ultrasound engineer has to increase the acoustic energy per transmit burst.

One approach is that the engineer can »simply« increase the transmit voltage. This can have significant ramifications, since special high voltage integrated circuits will be required. This tends to increase the amount of the electronics in the probe housing, and hence the size of the probe can become quite large. In addition, the more electronics, the more heat that is generated. This then pushes the engineer towards using active cooling[1], which further enlarges the size, cost, and complexity of the matrix transducer. Such methods are often associated with the larger, transthoracic matrix transducers.

1 Active cooling refers to active methods of removing heat from the probe, often relying on pumping a fluid (such as water) down the transducer cable to a heat exchanger inside the probe housing. This is very similar to how water cooling works in a car engine. The counter approach is to use passive cooling, which relies on heat conducting materials to wick the heat into the patient, into the hand of the scanning clinician, or into the transducer cable.

Fig. 2.5 The 3D reconstructed view of room and bookshelf

Fig. 2.6 Broad beam (compare to focused beam shown in Fig. 2.1)

The more electronics, the more heat is generated in the probe housing.

Alternatively, more acoustic energy can be transmitted by optimizing the transfer of electrical energy into acoustic energy. This could include the use of single crystalline piezoelectric materials and the use of advanced matching layer materials that facilitate the transfer of sound energy from the acoustic stack into the human body. Because of size and heat constraints, such methods of efficient design are most applicable to transesophageal matrix transducers.

2.1.3 Fully sampled matrix array transducers

Over the last 30 years, numerous ultrasound methods have been used to scan 3D volumes in the body, although with minimal clinical or market success (▶ Chapter 1). Clearly there was a need for a matrix phased array transducer, particularly given the success of phased arrays over mechanically swept transducers for routine 2D applications.

In the late 1980s, such a matrix transducer was considered unfeasible. Even a small aperture transducer would require in excess of 2500 elements (50×50). The cable connecting the transducer with the ultrasound instrument would be extremely large, bulky, and expensive. Thus, the phased array beamformer (part of the ultrasound instrument) with its 2500 processing channels would be extremely expensive to manufacture.

This impetus drove the concept of sparse arrays. In sparse matrix arrays, only a small percentage of the 2500 elements are electrically connected or acoustically active. This sparse array technology was commercialized in the mid 1990s, but has since been abandoned by most manufacturers.

There were three major problems with sparse array technology:

- Although the elements were strategically placed to mitigate obvious grating lobes, there was still degradation in side-lobe performance, which would appear as degraded contrast resolution to the clinician.
- The reduced number of elements and the reduced area of the probe face associated with transmitting and receiving ultrasound data would result in a loss of signal-to-noise ratio. This would be seen as a clinical loss in penetration.
- Emergence of tissue harmonic imaging. During the 1990s, harmonic imaging quickly became mainstream for transthoracic cardiac imaging. In order to create a harmonic echo, transmit waveforms with sufficient pressures needed to be transmitted into the body. Sparse arrays, with only a fraction of the elements transmitting, would have been incapable of creating the mechanical index (MI) necessary to create a clinically viable tissue harmonic image.

Given the practical limitations of sparse arrays, there was clearly a desire to produce a »fully sampled« matrix array, where all of the transducer elements were capable of both transmit and receive. Yet the practical limitation of supporting a transducer and system with a very large element and channel count remained.

Given the practical limitations of sparse arrays, there was clearly a desire to produce fully sampled matrix array transducers.

This led to the development of micro-beamforming [10]. The key concept is to electrically group small arrays of elements (referred to as patches) and to coherently combine their output to a single wire going back to the »mainframe« beamformer. A typical patch for a cardiac matrix probe might contain approximately 25 elements configured in a 5×5 geometry. Assuming a 128 channel beamformer with 128 wires, connecting the transducer to the mainframe beamformer would provide a fully sampled array of 3200 elements.

In other words, micro-beamforming allows one to achieve the same uncompromised performance as a fully sampled array having 3200 wires and 3200 analog-to-digital converters, which can be achieved in a transducer with the size and cost appropriate for clinical use. A micro-beamformer acts, just as its name implies, like a very small electronic beamformer. Its sole purpose is to »point« its 25 elements (on both transmit and receive) toward the desired scan line.

In Fig. 2.7, a returning echo (wave field on the upper left) impinges on the face of the probe. For illustration purposes, there are only three elements in a patch. The sound arrives at the top element first and arrives at the bottom element last. The objective of each micro-beamformer is to time align all of the echoes from each element in its patch. Since the elements within a patch are co-located, the delay corrections can be quite small, and therefore the electronic circuitry in the matrix transducer can be small as well. Each micro-beamformer has its own output, which is wired back to the mainframe beamformer (back to the ultrasound system through the transducer cable assembly; Fig. 2.8). The roles of the mainframe beamformer are to time align the patches and to produce the parallel receive beams for each transmit beam.

There is a common misconception that a given acoustic scan beam is matched to a single element. Although a matrix array of elements might look like the CCD (digital film) of a camera, the matrix elements are better envisioned as the camera lens (which is used to focus the light).

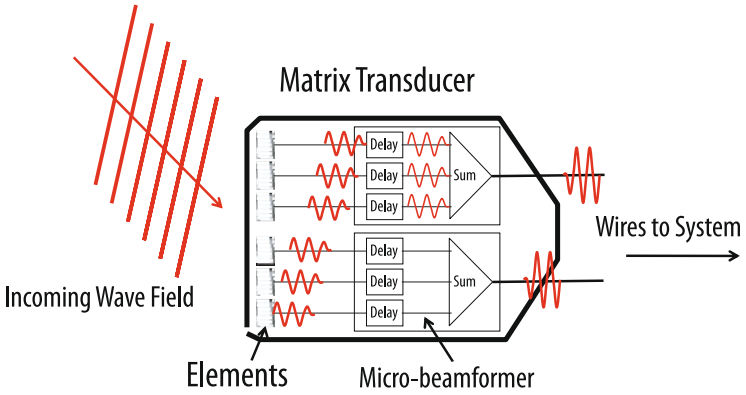

Fig. 2.7 Illustration of the principle of micro-beamforming

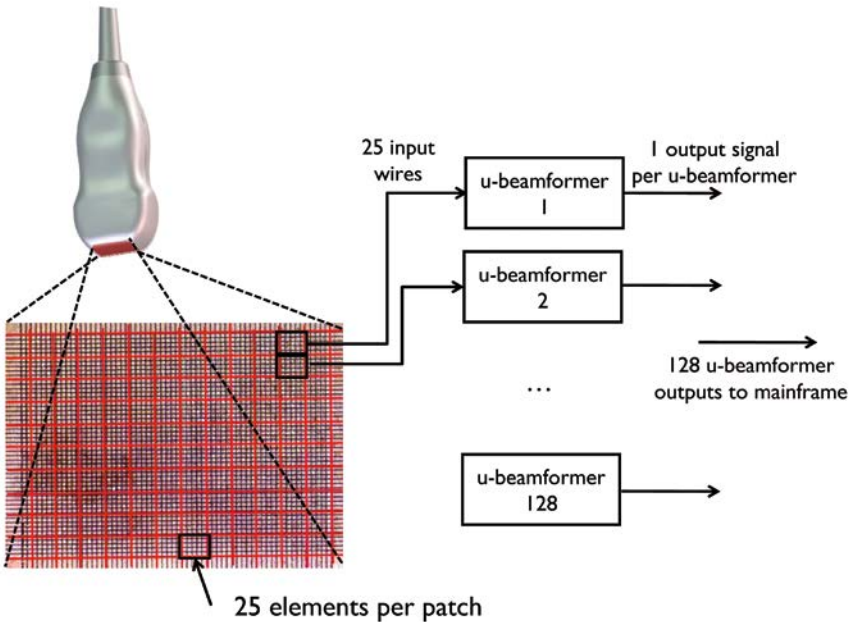

Fig. 2.8 Illustration of connection between micro-beamformers and mainframe beamformer

2.1.4 Parallel receive beam processing

One of the key ways to improve frame rate in current 3D ultrasound machines is through the use of parallel receive beam processing. Using the flashlight analogy will help illuminate how this works. In the darkened room, assume that there are four observers, but only one flashlight. The flashlight is slightly defocused to increase the spot size to 20 cm in diameter (it was 10 cm in the original scenario). Now each observer is asked to focus on a different quadrant of the light beam: observer A is asked to focus on the top left, observer B on the top right, C on bottom left, and D on the bottom right. So now, with each illuminated section of the wall, four smaller regions can be simultaneously interrogated (Fig. 2.9).

The use of parallel beam processing is one key way to improve frame rate in 3D ultrasound machines.

Most modern 3D phased array ultrasound machines use four or more simultaneous receive beams (the observers) for each transmitted beam (the flashlight). However, each receive beam shares the same aperture (probe face), the same elements, the same patches, and the same micro-beamformers. By using slightly different time delays in the mainframe beamformer, as applied to the micro-beamformed output, allows the »mainframe« beamformer to steer the receive beams in slightly different directions from one another.

2.1.5 Three-dimensional color flow

Unfortunately, 3D color flow inherits all of the B-mode issues (as discussed previously) and then compounds it with its own problems. Clearly the largest problem is frame rate, which can be experienced in 2D color flow. Why is color flow so much slower than B-mode?

In order to observe Doppler velocity shifts from blood, color flow imaging requires that an »ensemble« (a group) of scan lines be fired in the same direction. This is required for two reasons. The first is the need of a »wall filter«, which is used to remove the stationary tissue echoes that can be thousands of times larger than the

Fig. 2.9 Illustration of principle of parallel receive beam processing using the analogy of the flashlight

blood echoes. In its simplest embodiment, a wall filter can simply subtract the echoes from consecutive scan lines. Echoes that have not moved (from tissue) will be identical in both scan lines and, hence, will be eliminated. Moving echoes (from blood) will not be subtracted, since they are displaced scan line to scan line.

Clearly the largest problem of 3D color flow is frame rate.

The second justification for having an ensemble of color flow lines is to detect the motion. By observing the time shift in the blood echoes (post wall filter) from one scan line to the next, the engineer can directly calculate the displacement distance. Dividing by the pulse repetition interval (PR=time between subsequent scan lines=~0.2 ms) will calculate the velocity, which is then encoded as a color and displayed. Note that this simple explanation can also be used to derive the Doppler equation.

Based upon the prior discussion, the smallest theoretical ensemble is three. However, for numerous practical reasons, a typical ensemble length for 2D color flow consists of 6–10 or more scan lines (for commercially available instruments). Recalling the 250 transmit events available for a real-time B-mode volume, a real-time color volume would be further limited to only 25 transmit scan directions (assuming an ensemble of ten acoustic lines per scan direction). Even at the relatively coarse transmit spacing of three degrees, the real-time 3D color would be limited to a 15×15 degree volume (lateral × elevation) using a 5×5 scan grid. This is a very small volume for routine clinical use.

There are four options available to the ultrasound engineer to make this 3D color flow volume larger:

- Coarsen the transmit line spacing. Not only will this adversely affect the color flow resolution (more lateral/elevation smearing of the color data), it will also degrade sensitivity and the ability to detect pathologic flows.
- Decrease the number of acoustic lines within an ensemble. This will also degrade color flow sensitivity.
- Sacrifice frame rate.
- Sacrifice real-time and use gating: acquire smaller subvolumes over multiple cardiac cycles.

Again, all of these techniques are used to varying degrees by the various 3D manufacturers.

2.1.6 Principle of depth volume rendering for 3D imaging

Volume rendering is a method by which a 3D matrix of numbers is composited to produce a 2D image. Although the resultant 2D image looks very similar to the 2D image produced by surface rendering (the method common to the video gaming industry), the source data and the method of compositing can be quite different.

Volume rendering is very analogous to the way in which the human eye sees the 3D world. For each rod (or cone) on the retina, a »ray cast« originates from the eye such that the light intensity on the rod will be dictated by the first opaque structure intersecting the ray cast.

In volume rendering, a »2D plane« is spatially oriented with respect to the 3D object. The 3D object (or volume) contains samples at each point in the XYZ space. Note that 3D samples are referred to as voxels, and 2D samples as pixels. For each pixel on the 2D plane, a ray cast line is created which is perpendicular to the 2D plane and which intersects the 3D object. This is shown in Fig. 2.10. In 3D echocardiography, the voxels could either be B-mode magnitudes or color velocities as acquired from a matrix transducer.

Each sample along the ray cast, corresponding to the distance from the pixel, will be assigned the value from its intersected voxel. This is shown in Fig. 2.11. Whereas the ray cast signal looks noisy, this is typical of ultrasound data. The high frequency variations correspond to the speckle variations seen in clinical images, which make volume rendering more difficult.

Compositing is the process that converts ray cast values into the single value assigned to a pixel. For the human eye analogy, compositing effectively assigns the light from the first opaque structure to the rod. This occurs because each voxel in 3D space has both reflective and translucent physical properties. For example, the ray cast will sample »air« voxels, but these are not visualized since they are 100% translucent.

Volume rendering of ultrasound voxels is a bit more difficult. Ultrasound voxels are not inherently translucent or opaque. Therefore, during the compositing process, the algorithm will need to calculate an opacity for each ultrasound voxel. In the case of B-mode anatomical data, it would be desirable to make the heart chambers and the blood pools translucent. Since such echoes tend to be hypo-echoic, one can use the amplitude of the B-mode data to assign the opacity. This often involves a threshold such that any ray cast value below the threshold will be considered 100% translucent. This threshold is illustrated by the red line in Fig. 2.11.

In a very simple compositing implementation, one could simply select the first sample above the opacity threshold. In Fig. 2.11, the first sample above the threshold has a value of 29, which would then be assigned to the 2D pixel. Repeating this operation for every pixel in the 2D plane would produce the volume rendered image similar to the one of the right in Fig. 2.10. Actual volume rendering in commercial 3D ultrasound machines is a bit more complicated. Further insights can be gained from [11].

Recent advances in 3D visualization have involved the use of colorization, or chroma, to imbed depth information in the

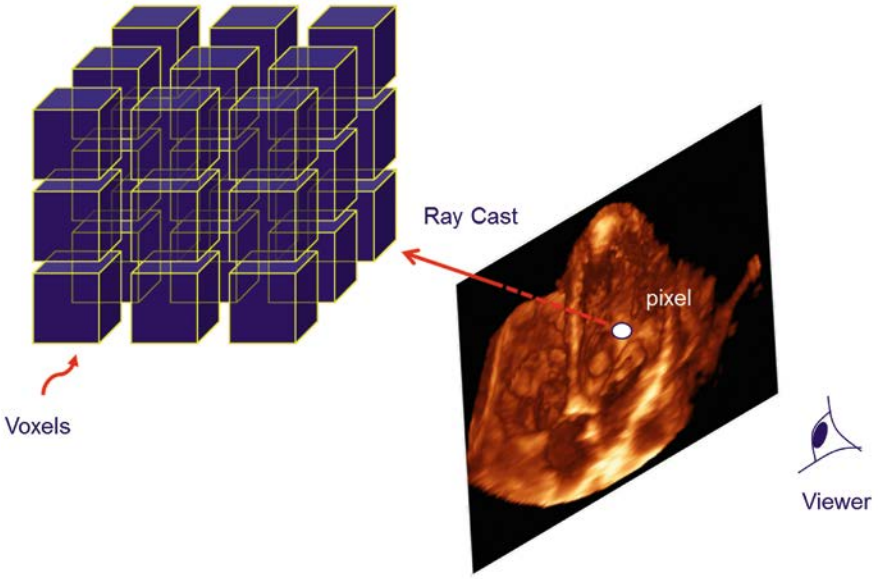

Fig. 2.10 Volume rendering

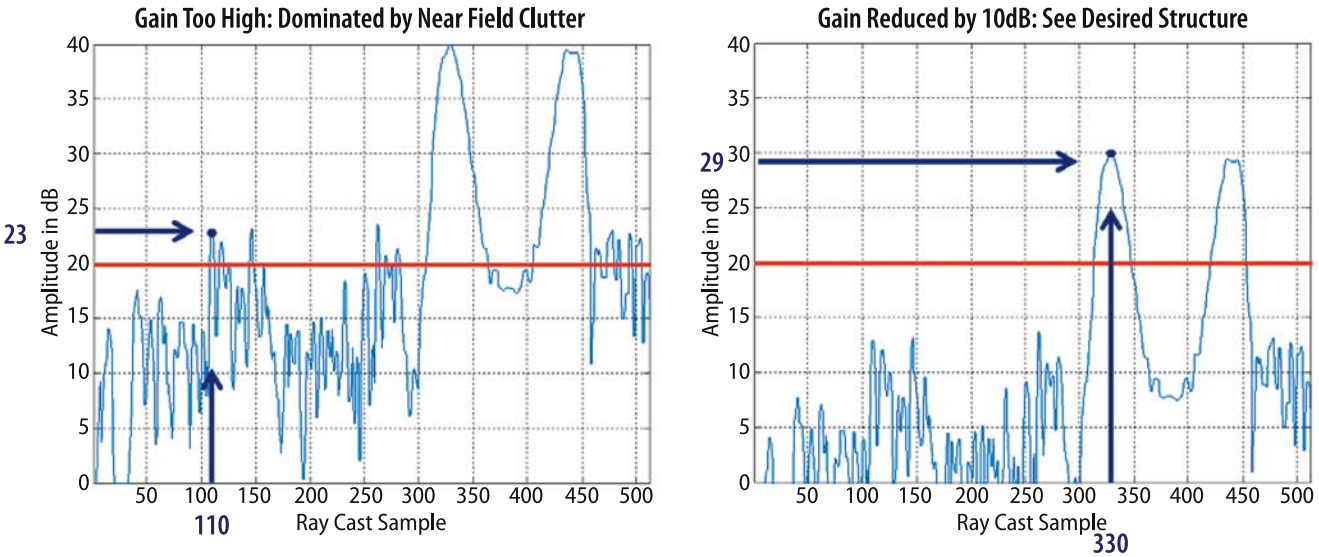

Fig. 2.11 Samples along a ray cast corresponding to a single pixel. The horizontal axis corresponds to the sample index, or to the distance from the pixel. The vertical axis corresponds to the amplitude of the B-mode echo. *Left* Illustration of a situation where the gain is too high, such that near field clutter exceeds the opacity threshold (*red line*). *Right* The gain is reduced, such that the desired anatomical structure (@ sample 330) is now visualized. Note the deeper anatomical structure located at 440. This will not be observed because it is blocked by the first structure at 330

volume rendered image [12]. This helps to increase the 3D depth perception of the image without having to resort to stereoscopic methods (e.g., 3D glasses).

Internal to the ultrasound instrument, a depth-weighted compositing algorithm will calculate two values for each ray cast line: the amplitude of the first opaque sample and its depth location. Referring to the right image in Fig. 2.11, the first opaque anatomical structure has an amplitude of 29 with a depth index of 330. From these values, two separate images are created: the first being the classic volume rendered image, which has already been described, and the second being a depth image (lower left in Fig. 2.12). The intensity of the depth image corresponds to the depth of the sample; shallow structures (close to the viewer) are shown in black, whereas deeper structures are shown as white.

These two internal images will then be combined through a 2D look up table (LUT). It has the special properties that the intensity of the RGB (red, green, blue) colors will only be dictated by the value of the composited value, and that the hue of the RGB colors will only be dictated by the depth index. Repeating this operation for every pixel in both source images will produce the depth-rendered image as seen on the right in Fig. 2.12. This allows twice the information to be communicated in a very intuitive manner.

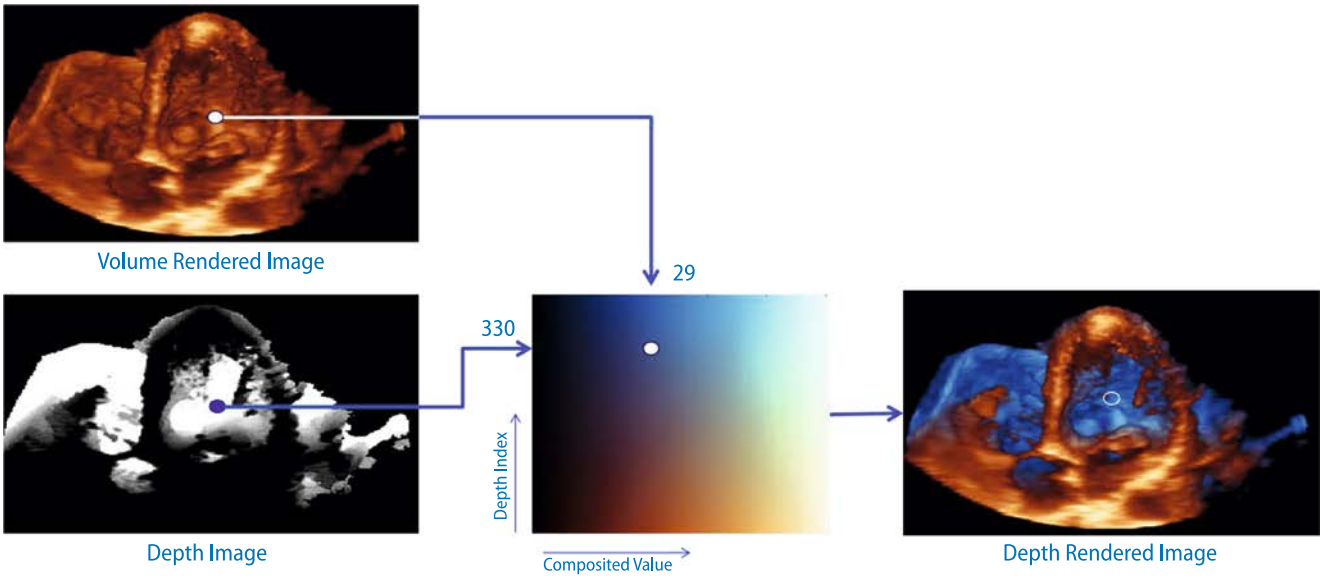

Fig. 2.12 Volume rendering using depth-weighted colorization

2.2 The probes

Today, matrix array probes exist for both transthoracic and transesophageal live 3D scanning. Compared to the first commercial matrix array probe by Volumetrics (▶ Chapter 1), the transthoracic probes have decreased significantly in size and nowadays are hard to differentiate from conventional 2D phased array probes (Fig. 2.13 and Fig. 2.14). There were several challenges to overcome in building the latest matrix array transducer generation (▶ Section 2.1). Because of the large number of ultrasound elements (2400–2500, compared to 64–128 elements in 2D phased array probes), the size of the transducer aperture became too large to scan through the narrow intercostal space. Decreasing the size of ultrasound elements and narrowing the field of ultrasound elements, however, caused increased problems with heating which have been successfully solved in present 3D transthoracic and transesophageal transducers by using more efficient piezoelectric elements and active or passive cooling techniques. Other difficulties to overcome pertained to the data transfer of the analogue high-frequency (HF) signals from the transducer via the transducer cable to the digital signal analysis unit within the ultrasound system. As mentioned in ▶ Section 2.1, computer boards for signal analysis nowadays have been miniaturized to fit into the transducer handle, which results in a relatively thin and flexible transducer cable for digital signal transfer to the ultrasound system for further analysis and display. We can expect that 3D ultrasound systems will be further miniaturized in the near future, and even cordless transducers using wireless transmission are conceivable.

In the newest transesophageal and transthoracic 3D probes, all ultrasound modalities have been successfully integrated.

As a practical limitation, it should be noted that some 3D probes are still currently limited to 3D imaging only and do not provide standard 2D imaging modes and spectral Doppler acquisition. This is particularly the case in some transthoracic probes, thus, urging the examiner to change transducers during 2D and 3D scanning, whereas in 3D transesophageal probes all ultrasound modalities – 2D, 3D as well as color and spectral Doppler – have been successfully integrated as an essential requirement. The newest generation of transthoracic 3D probes combines all features of previous matrix-array 3D probes and standard 2D probes (Fig. 2.13). Such all-in-one-probes allow full integration of 3D imaging into a standard transthoracic echocardiographic workflow, thereby overcoming the need to change between 3D and 2D transducers. This new all-in-one matrix array transducer technology is also capable to provide full volume acquisition in one beat (one-beat full volume mode) as well as live 3D color Doppler imaging (Fig. 2.15).

2.3 Live 3D echocardiographic examination

In principle, there are two ways of performing a RT3D echocardiographic examination in clinical practice:

1. to perform a standard 2D echocardiographic examination and to add RT3D datasets of selected structures or regions or
2. to perform a full RT3D echocardiographic examination with standard 3D views and to add only spectral Doppler datasets with information that cannot be obtained by RT3D datasets [13](14].

Currently, the first protocol is certainly more common for several reasons. Most echocardiography laboratories are still on a learning curve and not experienced enough to fully rely on RT3D datasets yet. In a recent first recommendation paper on 3D image acquisition and display, however, prerequisites for the integration of 3D echocardiography into routine patient examinations very

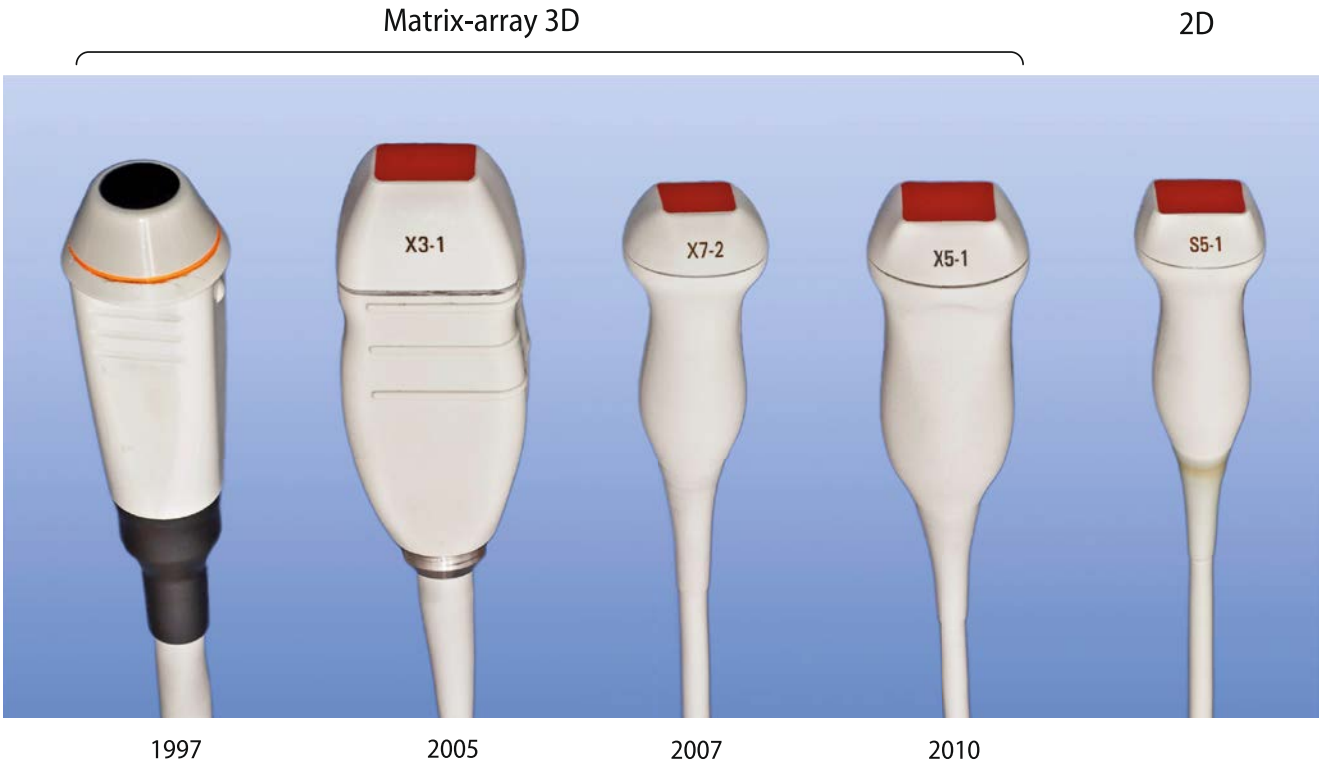

Fig. 2.13 Evolution of transthoracic real-time 3D matrix array transducers: (from *left* to *right* with year of first release). The first commercially available matrix array transducer with a sparse array of 50×60 elements and a very large transducer cable (1997, Model 314U, 3.5 MHz, Volumetrics Medical Imaging, USA) [20) (21]. The adult matrix array transducer with 2400 active elements with a relatively bulky handle due to embodied postprocessing electronics, but a relatively thin transducer cable (2005, Model X3-1, 1–3 MHz, Philips Medical Systems, USA). Further miniaturized pediatric matrix array transducer with 2500 active elements (2007, Model X7-2, 2–7 MHz, Philips Medical Systems, USA). The new all-in-one adult matrix array transducer with an ergonomic sized handle which also operates with all common 2D imaging and spectral Doppler modes (2010, Model X5-1, 1–5 MHz, Philips Medical Systems, USA). *Far right* A conventional 2D phased array transducer with 80 elements is shown (Model S5-1, 1–5 MHz, Philips Medical Systems, USA)

Fig. 2.14 Side-by-side illustration of a transesophageal live 3D matrix array probe with 2600 elements (*top*; Model x7-2t, 2–7 MHz, Philips Medical Systems, USA) and a standard multiplane 2D phased array probe with 64 elements (*bottom*; Model Omni 3, Philips Medical Systems, USA)

outlined as to be (1) a single transducer capable of 2D and 3D imaging, (2) accurate automated chamber quantification, and (3) automated display of standard 3D and 2D echocardiographic views [15]. With the latest all-in-one 3D matrix array probes the first requirement has been achieved. Although, analysis of 3D datasets can be time consuming, several semiautomated 3D quantification tools have been developed or will be available in the near future which addresses the second and third requirements. However, the recent recommendations on 3D image acquisition and display provide a first consensus on the use of 3D echocardiographic techniques in clinical routine practice, thus, paving the way for more standardized 3D examinations and the broader use of RT3D echocardiography. Besides the need of national and international recommendations, two other factors are important for wider clinical application of RT3D echocardiography: (1) expert-guided training courses to ease and accelerate climbing the learning curve and (2) the manufacturers to make the handling and analysis of 3D data as feasible as 2D data. In other words, the workflow of integrating 3D echocardiography into the examination needs to be made easier.

There is still some uncertainty about how to integrate 3D echocardiography into the patient examination that needs to be overcome.

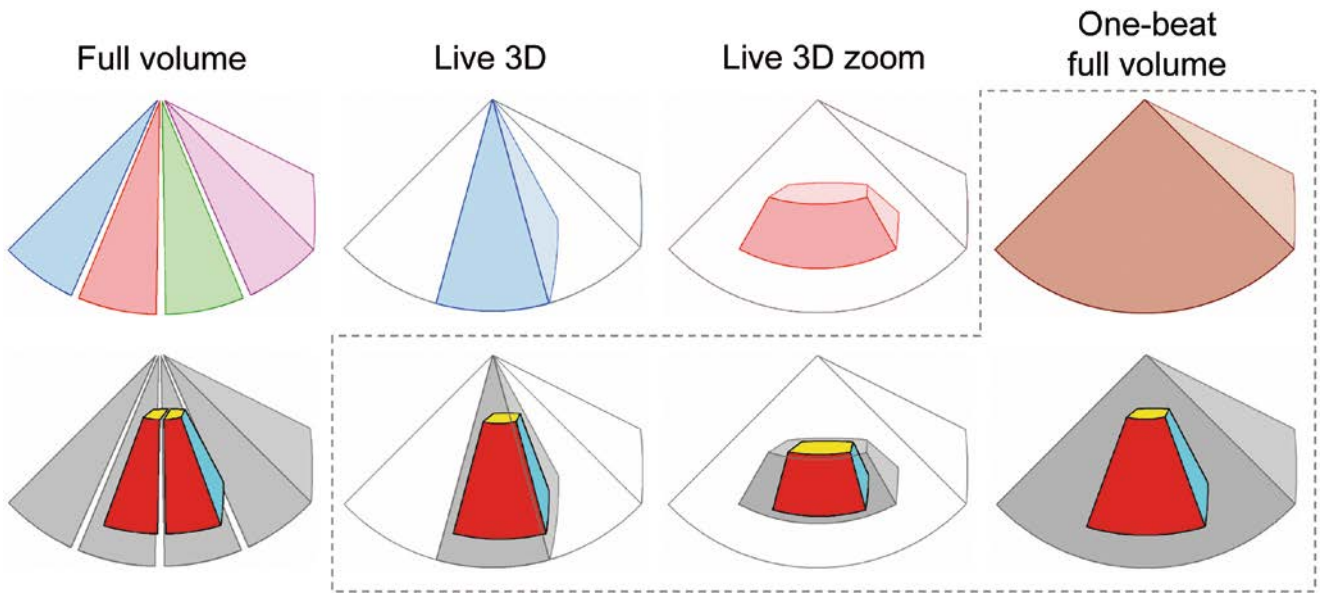

Fig. 2.15 Schematic representation of different 3D volume acquisition modes. *Top row from left to right* Full volume composed of four or seven subvolumes; live 3D mode; live 3D zoom mode; one-beat full volume. *Bottom row* The respective 3D volume acquisition modes with additional 3D color Doppler volumes are illustrated. The volume modes within the dashed line – one-beat full volume with and without color Doppler and live 3D color Doppler – are not available in all 3D systems yet. The 3D datasets included in this book were all acquired using the four 3D volume acquisition modes outside the dashed line

2.3.1 First steps and learning curve

When RT3DE with live 3D visualization became widely available for the first time in 2002 (after its introduction at the European Society of Cardiology Congress in August 2002), the majority of users discovered that the acquisition of transthoracic datasets (and later in 2008 similarly with transesophageal datasets) was feasible and relative easy to learn. However, further analysis of 3D datasets, either onboard or on an external workstation, required special knowledge and skills of 3D dataset cropping, 3D orientation, and knowledge of how to use the 3D analysis software. This was particularly important as 3D datasets often reveal their entire information only after cropping and slicing, like in a crystal geode (▶ Section 2.3.2, »Multiple-beat full volume mode«) and unlike 2D images that instantly show their information as it is acquired. Beyond this, 3D analysis is still considered to be more time consuming compared to conventional 2D image analysis. As a consequence, users acquired large numbers of 3D datasets without drawing diagnostic information from them and finally run the risk of becoming dissatisfied even before they started to move up the learning curve.

The 3D datasets often reveal their entire information only after cropping and slicing.

Learning from this experience, 3D manufacturers and clinical experts have offered interactive 3D training courses to help novice users to overcome difficulties when learning to use 3D analysis software for cropping, orientation, and displaying of 3D images [16]. Even more, measurements in 3D datasets, including size of an atrial septal defect (ASD), left atrial (LA) volume, or quantification of global and regional left ventricular (LV) function as well as the more advanced quantification of mitral valve anatomy required initial step-by-step interactive demonstration.

2.3.2 Three-dimensional acquisition (modes and image settings)

Nowadays, 3D data acquisition, in principle, can be easily integrated into standard transthoracic and transesophageal echocardiographic examinations by using an all-in-one matrix array transducer capable of 2D, spectral Doppler and 3D imaging and switching to 3D modes by pressing a button [17]. As mentioned above, different modes of 3D volume acquisition are available. Currently, the most common modes are live 3D, live 3D zoom, full volume as well as full volume color Doppler acquisitions, which are available on both transthoracic and transesophageal matrix array probes (Fig. 2.15).

However, as the diagnostic applications of live 3D in transthoracic echocardiography (TTE) and transesophageal echocardiography (TEE) are different, it is of critical importance to understand the characteristics of the different volume modes in order to choose the most appropriate one for a given setting, which is analogous to choosing the appropriate gear when driving a car instead of always driving in the same gear. Basically, we can currently state that real-time 3D TTE is more suitable for whole heart capture by gated or one-beat full volume acquisition for quantitative 3D analysis of global and regional LV function, RV volumes, and LA volumes, whereas real-time 3D TEE is especially suited for detailed 3D imaging using zoomed live 3D volume acquisitions for the analysis of valves, defects, and other structures, e.g., cardiac masses [13]. The different real-time 3D volume modes can be characterized as described below. As has been previously mentioned, there are a number of important interactions between the physical characteristics of a 3D dataset; these will now be described in more detail.

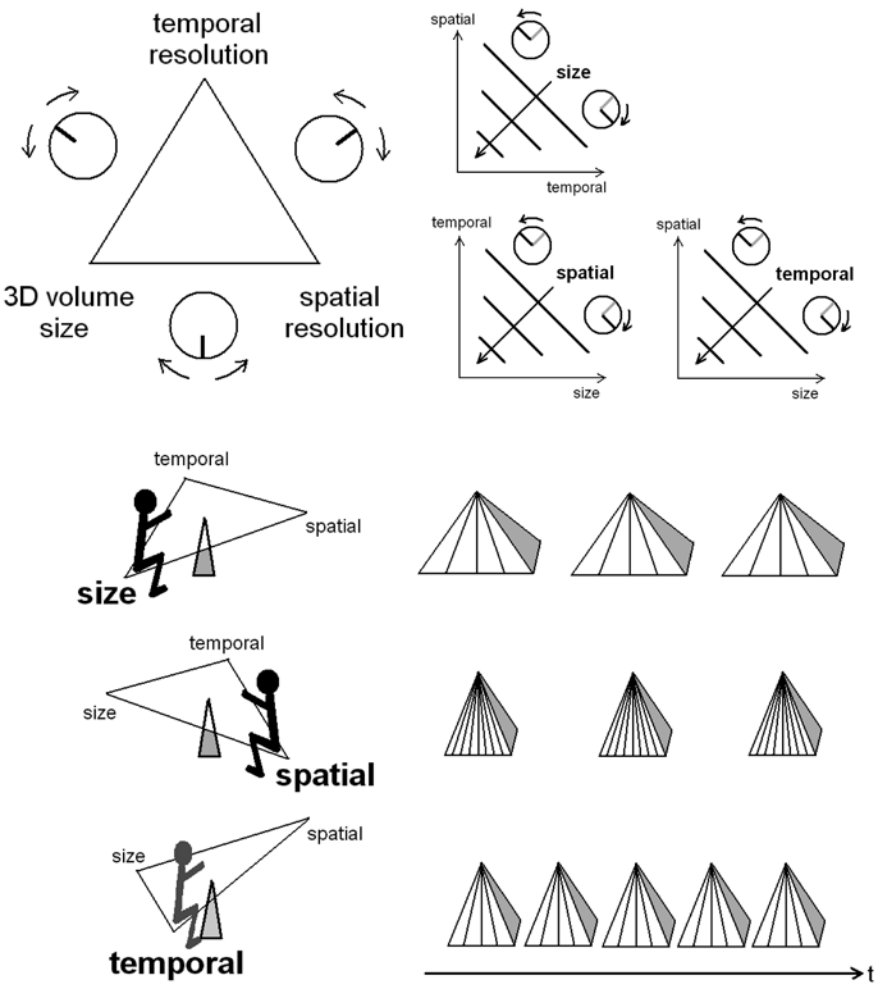

Fig. 2.16 Principle of interdependency of 3D volume size, temporal and spatial resolution. *Upper left* The interdependency is expressed by a triangle where all three parameters are connected to each other. The dial knobs between each pair of parameters indicate the option for weighting more to the one or the other. *Upper right* The three diagrams illustrate the effect of the dialing for all possible settings of the three parameters. The *upper diagram* shows for a given 3D volume size the effect of dialing more towards increased spatial resolution causing lower temporal resolution and vice versa. With increasing 3D volume size (in direction of the size arrow), however, both temporal and spatial resolution have to decrease. *Lower part* Interdependency between the three parameters is illustrated by the triangle balanced on a center axis point. The preference set to a large 3D volume size (*top*) causes lower temporal resolution (as indicated by the separation between pyramids on the right) and lower spatial resolution (as indicated by the three wide-spread lines). The preference set to higher spatial resolution (*middle*) causes smaller 3D volume size at a lower temporal resolution (the higher spatial resolution indicated by a larger number of lines with higher density). The preference set to higher temporal resolution (*bottom*) causes smaller 3D volume size and lower number and density of lines

The triangle of interdependency between temporal and spatial resolution and volume size

Setting up real-time 3D volumes is fundamentally determined by a triangle of three interdependent characteristics as these are (1) the volume size, and here particularly the width in lateral and elevation dimension, (2) temporal resolution, and (3) spatial resolution (Fig. 2.16). As there is only one transmit beam travelling through the scan volume of a given size and the velocity of the ultrasound beam is a constant of approximately 1540 m/s, the time required for the beam to fully travel through the entire volume is determined by two factors (1) the size of the scan volume and (2) the line density which determines the total number of scan lines in the scan volume as already emphasized in the technical part in ▶ Section 2.1.1. As the line density determines the spatial resolution of the 3D voxel dataset and the resulting 3D image and the total number of scan lines required to scan the volumes determines the temporal resolution the following consequences and trade-offs exist:

- **For a larger 3D image** increasing the scan volume has the consequence of increasing the number of scan lines resulting in a longer time to scan the volume and, thus, lower temporal resolution or the consequence to accept a lower line density resulting in lower spatial resolution or a combination of both consequences (in newer systems this issue has been especially addressed by adding a dialing function that allows for weighting the trade-off to be more on temporal or more on the spatial resolution).
- **For higher image resolution** increasing spatial resolution by increasing the line density has the consequence of decreasing the size of the scan volume or to increase the number of scan lines required to scan the volume resulting in a lower temporal resolution or a combination of both consequences.

- **For higher temporal resolution** increasing temporal resolution by decreasing the number of scan lines required to scan the volume has the consequence of decreasing the size of the scan volume or to decrease the line density resulting in a lower spatial resolution or a combination of both consequences.

From these interdependencies and trade-offs, the following characteristics and practical recommendations for setting up 3D volume acquisition evolve.

One-beat 3D volume acquisition mode

In principle, in the latest generation of 3D systems one-beat 3D volume acquisition mode providing live 3D imaging can be set up to acquire volumes of almost any size from relatively narrow pyramid-shaped volumes and zoomed volumes up to large pyramid-shaped full volumes of more than 90° angle width. However, live 3D imaging with satisfactory temporal resolution requires a minimum of approximately 10–12 volumes per second. And therefore according to the interdependencies discussed above one-beat 3D volumes or live 3D volumes should be limited to a size that still provides a volume rate of at least 10–12 volumes per second and acceptable spatial resolution. For example, setting up a transthoracic one-beat full volume to a size large enough to encompass the entire left heart or even left and right heart temporal resolution in most 3D systems will drop to less than 10 beats per second or/and spatial resolution will drop such that satisfactory image quality is prevented. To address these requirements, for example, in some older 3D systems, a live 3D volume acquisition mode with a fixed narrow volume size was configured: angle width was about 30° in the elevation dimension and about 65° in longitudinal (azimuth) dimension (◘ Fig. 2.15).

Advantages of a live 3D volume of such narrow dimensions, however, are high spatial resolution (because of high line density) and high temporal resolution (because of a relative narrow scan volume). Therefore, such a narrow live 3D volume is particularly useful for visualization of structures that are small enough to fit into the narrow volume, e.g., small valves, small masses, or vegetations, which become more difficult to fit into a narrow live 3D volume the closer the structure is to the transducer. However, because in transesophageal scanning the orientation of the aortic valve plane is nearly parallel to the ultrasound beam direction, it fits very well into this narrow live 3D volume where it is located in an upright position (▶ Section 2.3.3). Compared with this, normal-sized mitral valves are oriented in a more perpendicular position and are usually too wide or too close to the transducer to fit into a narrow live 3D volume. In addition, in transthoracic live 3D scanning, the mitral valve fits better into the narrow volume because of the larger distance from the transducer (◘ Fig. 2.17). Note, however, that for a given line density visualization of a structure (i.e., valve) at a lower depth (as in transesophageal scanning) results in an effectively higher line density as the lines had a smaller distance to diverge as compared to visualization of a structure at a higher depth (as more frequent in transthoracic scanning) resulting in an effectively lower line density as the lines had a longer distance to diverge (◘ Fig. 2.18). As a practical recommendation in order to minimize this effect, the volume depth should be set up to visualize the structure of interest rather in the center or the near field of the volume instead in the far field, in both transthoracic and transesophageal 3D scanning. The one-beat or live 3D volume mode (according to its name) provides live 3D viewing during scanning. However, because of the pyramid shape of the volume, direct viewing of valves, for example, requires extra cropping of the volume, particularly in transthoracic 3D datasets. Today, switching to 3D color Doppler in the live 3D and live 3D zoom modes is available in all 3D echocardiographic machines of the latest generation (◘ Fig. 2.15).

Live 3D zoom mode

Live 3D zoom or zoomed live 3D is a very practical mode particularly in transesophageal 3D scanning. It is similar to zoom modes in 2D scanning in that it allows one to manually define a volume of interest (◘ Fig. 2.15 and ◘ Fig. 2.17), which can encompass a specific region or structure. As an advantage opposed to pyramid-shaped live 3D volumes, this zoomed live 3D volume mode eliminates the need for cropping of the volume because the volume can be defined in a manner that provides direct viewing of the structures of interest, e.g., direct en face viewing of the mitral valve (◘ Fig. 2.17). However, the price to pay for the advantage of a live 3D visualization of a structure of interest (e.g., the mitral valve) in a volume wide enough to encompass the entire structure frequently is a relatively low temporal resolution with a volume rate of between 10 and 16 volumes per second depending on the width of the defined zoom volume and a lower spatial resolution due to the lower line density. The lower line density results in less detailed surface rendering of fine structures, e.g., chords or vegetations, which appear thicker compared to when a narrower live 3D volume or full volume mode are used (◘ Fig. 2.19). Despite inferior temporal and spatial resolution compared to 3D volume modes with smaller volumes or higher line density, live 3D zoom is probably the most practical and most frequently used 3D mode during transesophageal 3D scanning.

Multiple-beat full volume mode

The description »full volume« relates to the size of the pyramid-shaped 3D volume being large enough to encompass the full heart, reaching angles up to more than 100° in both the longitudinal (azimuth) and elevation dimensions. As described above one-beat full volumes are inherently limited by low temporal and low spatial resolution. In contrast to a one-beat full volume or other live 3D modes, a multiple-beat full volume does not consist of a single volume but of multiple 3D volumes stitched together (◘ Fig. 2.15 and ◘ Fig. 2.20). As a consequence, the multiple-beat full volume mode does not allow live 3D viewing, because the number of subvolumes is needed to be acquired over a corresponding number of heart cycles by electrocardiographic triggering, which also gives it the name »gated« 3D acquisition. As typical for large multiple-beat full volumes, pyramid-shaped datasets from a transthoracic approach look unspectacular from the outside as there is no information about the inside of the object, like in a crystal geode (◘ Fig. 2.21). By changing the settings for line density and number of subvolumes, the size of the full volume can by defined prior to acquisition. Because a multiple-beat

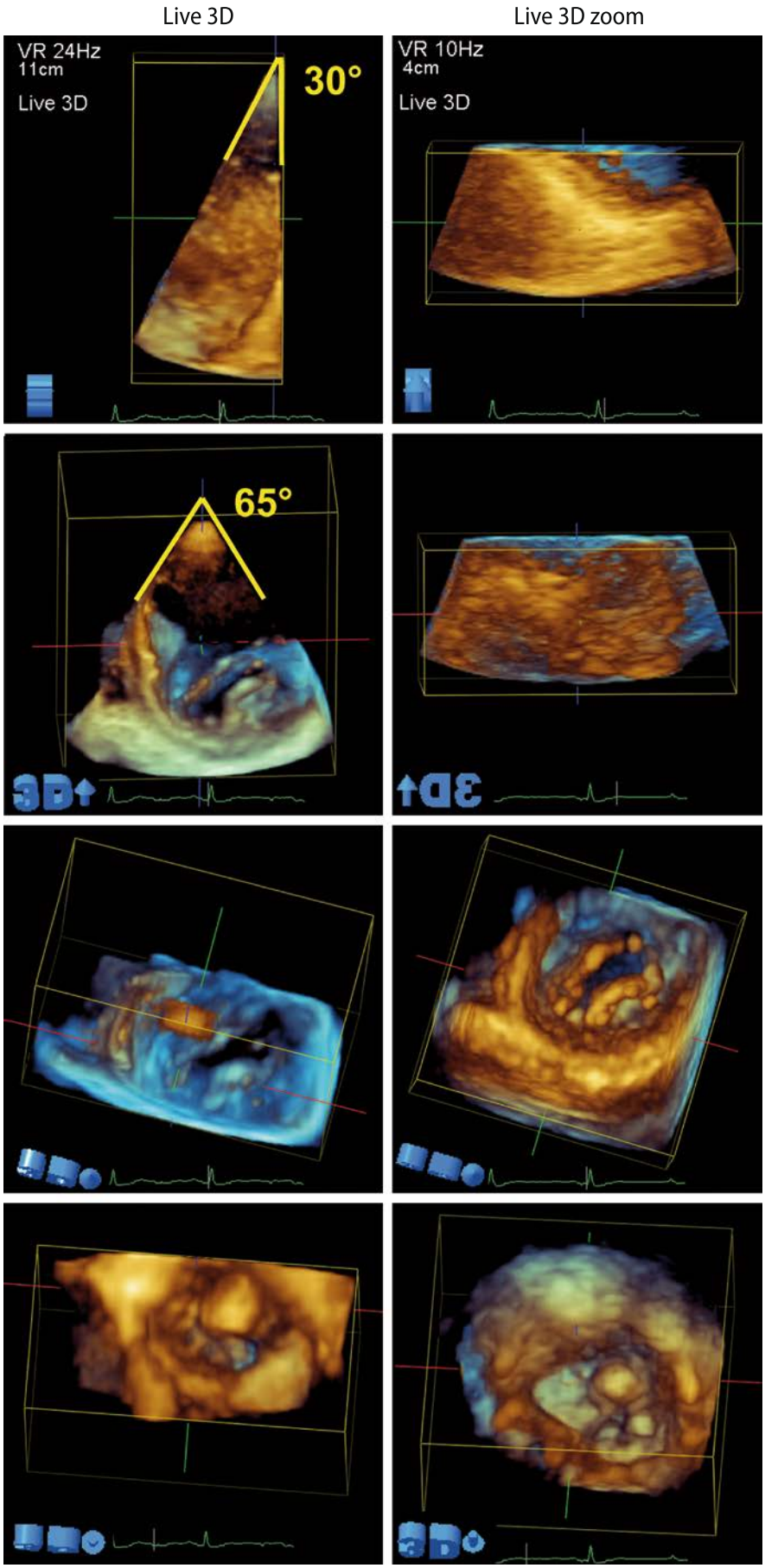

Fig. 2.17 Comparison of transthoracic live 3D mode (*left column*) and live 3D zoom mode (*right column*) in the same patient with mitral P2 prolapse. The two panels in the *top row* show side views of a live 3D volume with a narrow angle of 30° (*left*) and a wider live 3D zoom volume (*right*). The two panels in the *second row* show the same datasets from the other side after clockwise vertical rotation by 90°. The yellow crop box with the three axes (*red*, *green*, *blue*) helps with the orientation of the volume in space. An additional aid for spatial orientation is given by the *blue* **3D**↑ icon in the lower left corner of each figure. Panels in the *third row* show a view from the apex to the mitral valve: the live 3D volume provides a very narrow view to the mitral valve with signals of the left ventricular (*LV*) apex in between. The wider but flatter live 3D zoom volume provides a more direct view of the mitral valve. *Bottom* A view from the left atrium (*LA*) to the mitral valve with detection of P2 prolapse is shown. [→Videos 2.17A–H]

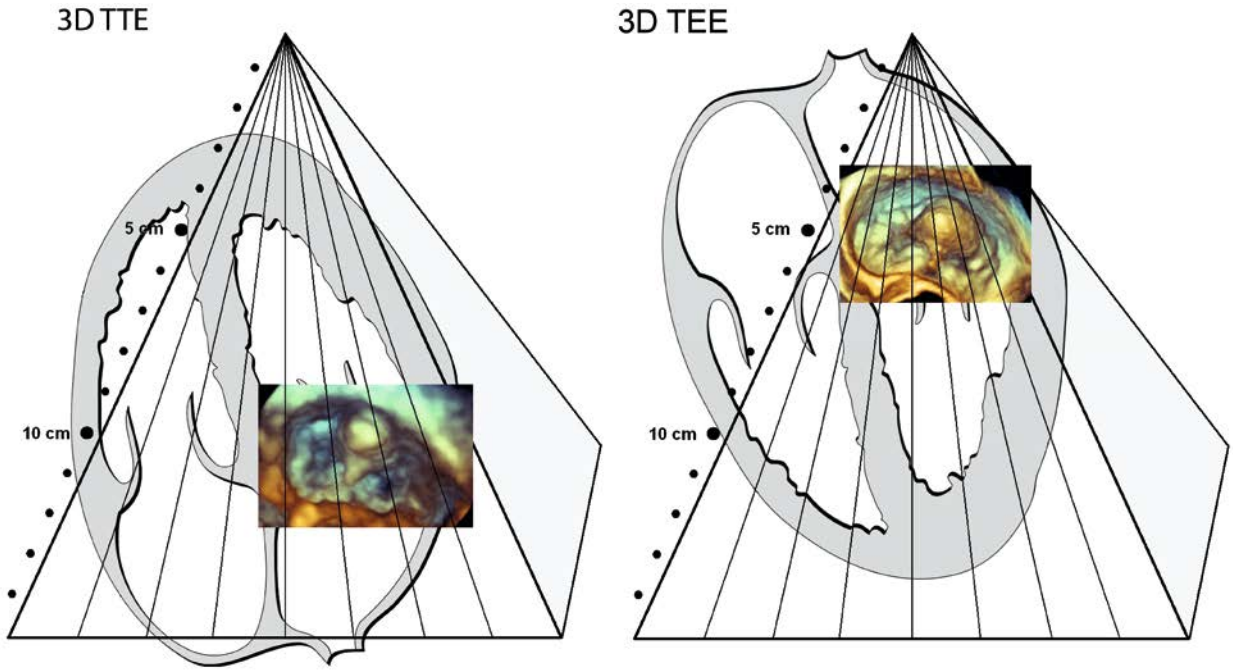

Fig. 2.18 Illustration of the relation between scanning depth, line density, and spatial resolution of 3D imaging from a transthoracic (3D TTE; *left*) and transesophageal approach (3D TEE; *right*). In the example of 3D scanning of the mitral valve, the schematic illustrates that due to the divergence of imaging lines with increasing depth spatial resolution and, thus, resulting 3D image quality is higher at a lower depth of 4–5 cm, which is where the mitral valve is located from a TEE approach (*right*), compared to imaging the mitral valve at a higher depth of 9–10 cm from a TTE approach (*left*)

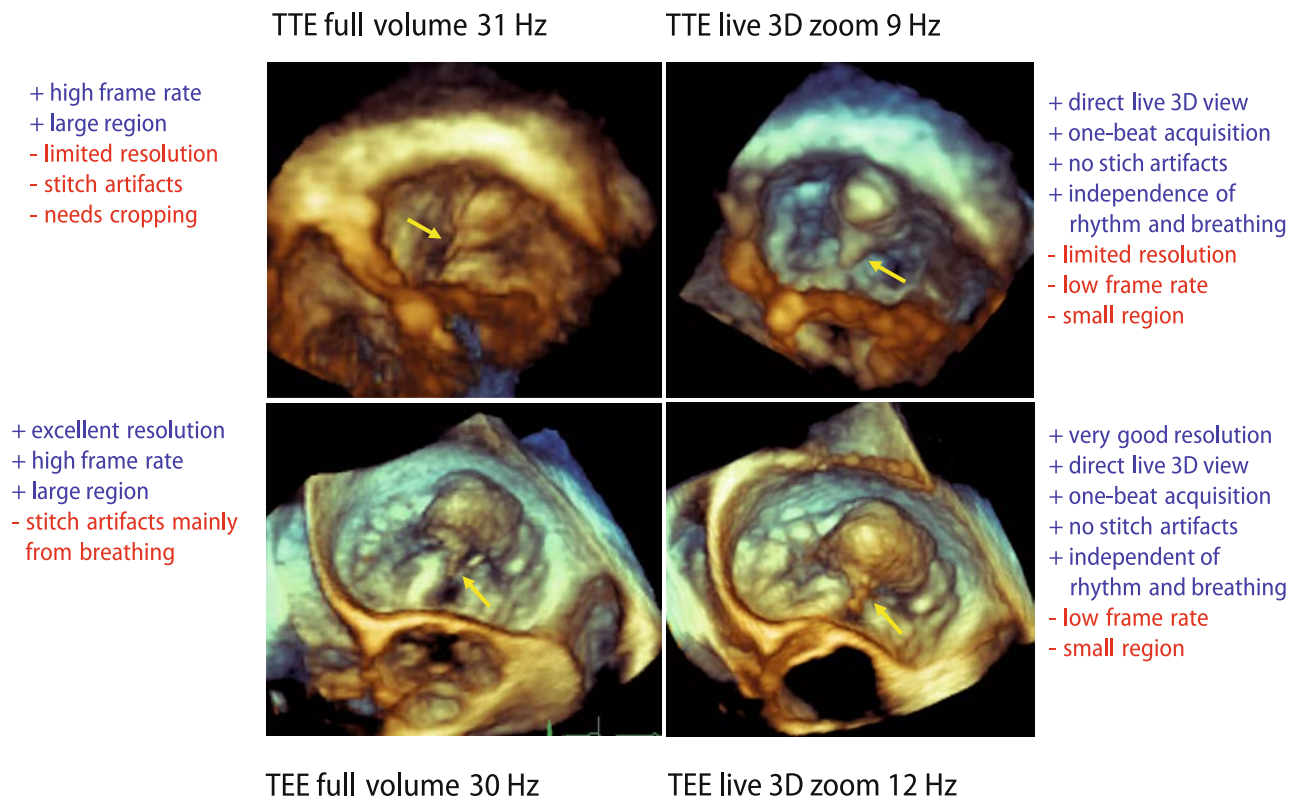

Fig. 2.19 Differences in image quality between transthoracic (*TTE*, *top row*) and transesophageal 3D imaging (*TEE*, *bottom row*) and full volume acquisition (*left column*) and live 3D zoom acquisition (*right column*) in the same patient are demonstrated. This shows an en face view from the left atrium to the mitral valve with P2 flail leaflet. For quick reference, positive (*blue*) and limiting (*red*) criteria of image quality are indicated for each 3D acquisition mode. Importantly, the different 3D acquisition modes have a significant impact on the representation of fine structures, such as a ruptured chord (*yellow arrows*) with the finest visualization of the chord in the 3D TEE full volume mode (*bottom left*) and the most blurred and thickened representation in TTE live 3D zoom mode (*top right*). [→Videos 2.19A–D]

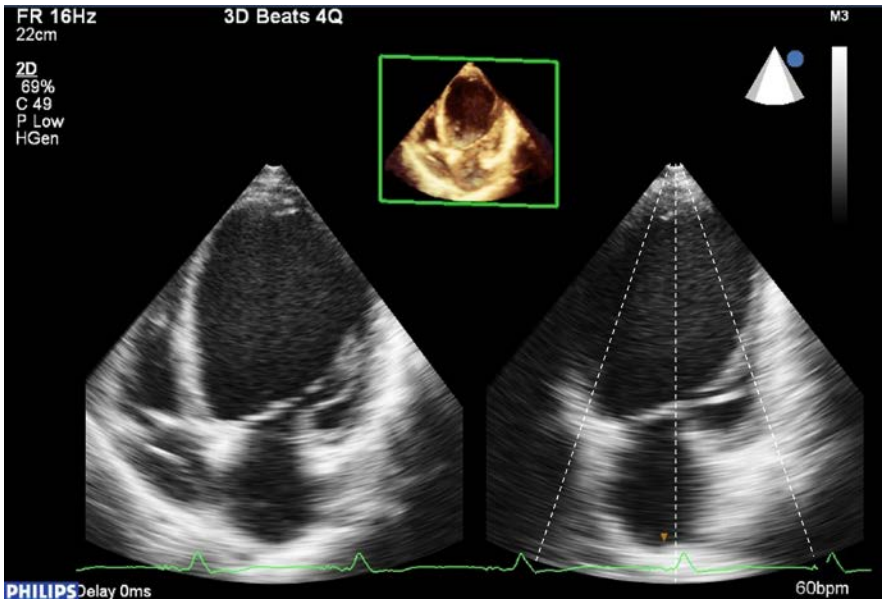

Fig. 2.20 Display during transthoracic four-beat full volume acquisition using a Philips X5-1 matrix array transducer. *Bottom left* Sector view represents the standard 2D scan plan according to the transducer orientation. *Bottom right* This view is a view perpendicular to the *left*, generated simultaneously by the matrix array probe in elevation dimension. The 3D image at the *top* represents the acquired 3D volume. *Lower right* The sector view is divided by *three dashed lines* dividing the four real-time 3D volumes stitched together

full volume consists of two, four or on some 3D systems even more 3D volumes stitched together, spatial and temporal resolution of the full volume dataset is as high as for the individual 3D volumes according to the trade-offs discussed above.

However, as a practical limitation, multiple-beat full volume datasets are prone to stitching artifacts (▶ Section 2.6) mainly caused by translation of the heart from breathing during the volume acquisition period. Therefore, stitching artifacts can be best prevented by breath holding in both transthoracic and transesophageal examinations or pausing respiration during cardiac surgery. If the length of the heart cycles vary, e.g., in atrial fibrillation, stitching artifacts can also result. In general, multiple-beat full volume is the most appropriate mode for transthoracic acquisition of large volumes encompassing the entire heart or the entire left ventricle for heart chamber quantification. In transesophageal examinations, the multiple-beat full volume mode allows acquisition of a wide 3D image encompassing multiple valves (Fig. 2.22) or large structures, e.g., large myxomas with good temporal and spatial resolution.

The full volume is the most appropriate mode for transthoracic acquisition to encompass the entire heart or the left ventricle.

3D color Doppler mode

In newer 3D systems, 3D color Doppler is available in both multiple-beat full volume acquisition mode and all one-beat or live 3D volume modes. In principle, as in 2D modes, a 3D color Doppler volume is superimposed onto the 3D volume with tissue information (Fig. 2.15) allowing to either turn-off color Doppler information or tissue information for visual assessment. Note that the fundamental trade-offs pertaining to volume size, temporal, and spatial resolution as discussed above are also valid for 3D color Doppler volumes. In addition, like multiple-beat full volume acquisition, multiple-beat 3D color Doppler volumes are potentially affected by stitching artifacts.

After acquisition of a 3D color Doppler volume, various color Doppler settings can be applied to the dataset significantly affecting the image quality. Important color Doppler settings include (1) color Doppler gain, (2) smoothing, (3) color vision, (4) filter, and (5) baseline velocity. Those settings can be either changes directly on the 3D system or off-line using analysis software such as Qlab (Philips) as long as the original 3D data format with raw data information is kept. The following settings are recommended for 3D color Doppler datasets acquired on a Philips 3D system: color Doppler gain =50, smoothing =2, color vision =2, filter =4–6, and baseline velocity =6 (default).

It is important to note that in 3D color Doppler acquisition, like in 2D color Doppler, depth setting has an important impact on temporal resolution in the way that the deeper the depth of the 3D color Doppler volume is set up, the lower the temporal resolution will be, whereas the depth setting in the normal tissue mode has little effect on temporal resolution or on one of the other interdependent factors as discussed above. As in 2D color Doppler, the autocorrelation process to determine color Doppler velocities along the color Doppler scan lines in a 3D volume is a time-consuming calculation, which results in longer scan time for each 3D volume when the depth of the 3D color Doppler volume and along with this the length of the color Doppler scan line increases.

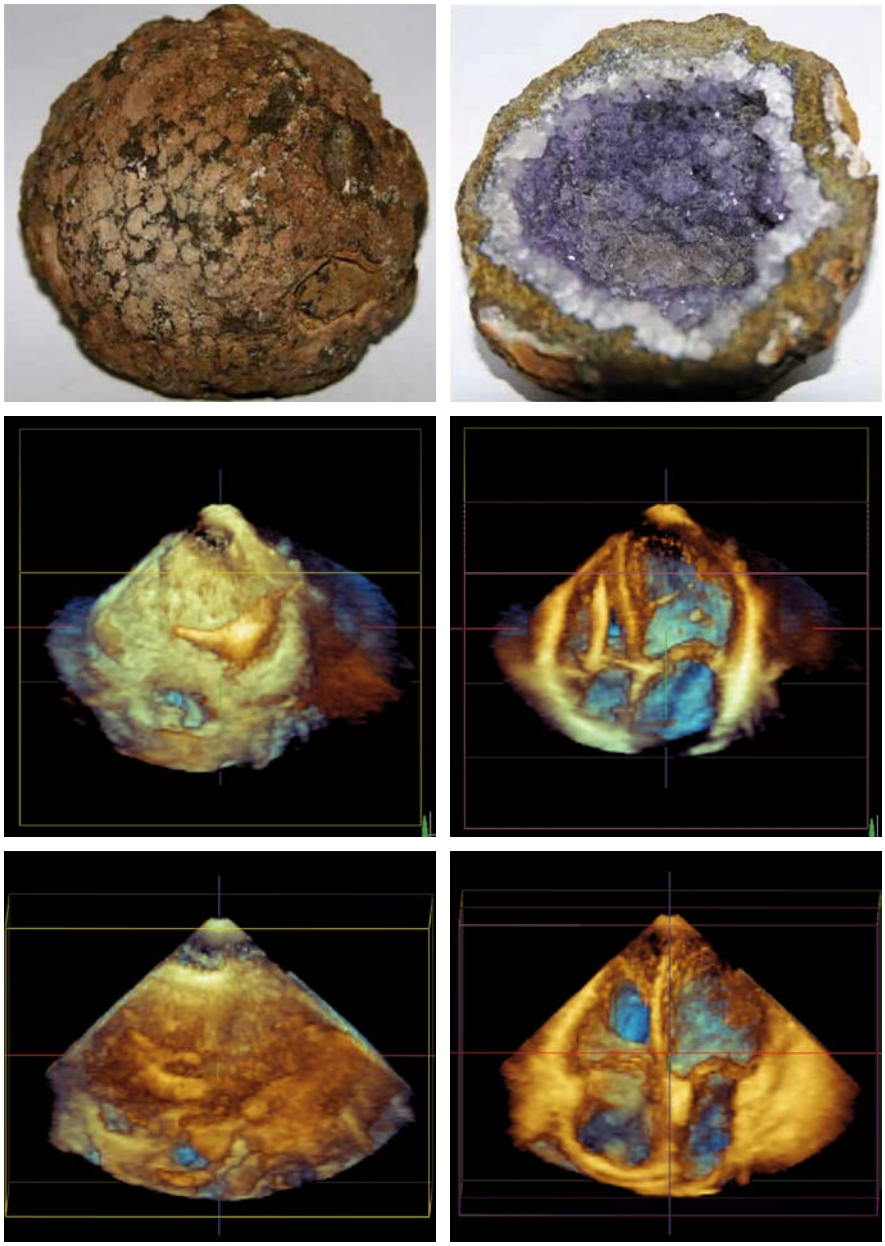

Fig. 2.21 Two examples of the full volume mode using TTE. The uncropped full volume (*left*) provides no information about the inside of the object, like a crystal geode (*top left*). Cropping the full volume towards its center provides a direct view of the important details inside. [→Videos 2.21A–D]

2.3.3 Standard 3D views and image orientation

Although live 3D echocardiography has been available since 2002, there were no unified recommendations on a standard examination protocol and standard views available until recently. This was partly because of the far more complex image information compared to 2D images. Now, a joint recommendation paper by EAE/ASE has been recently become available providing general information on how to set up 3D transthoracic and transesophageal examination protocols [15]. A first examination protocol for transthoracic RT3DE (RT3D TTE) with 3D image orientation according to anatomic cut planes of the heart in transverse, sagittal, and coronal orientation was proposed by Nanda et al. [18], where (1) a transverse cut plane orientation represents a vertical plane which is oriented almost perpendicular to the long axis of the heart and therefore provides short-axis views of the heart, for example a short axis of the LV, (2) a sagittal plane represents a vertical plane along the long axis of the heart that divides the heart into a left and right half, providing for example a long-axis view of the left ventricle, LVOT, and aortic root, and (3) a coronal plane represents a plane along the long axis of the heart but perpendicular to the sagittal plane, thus, separating the heart in an anterior and posterior half, providing for example a typical 4-chamber view (Fig. 2.23). More recently, a systematic characterization of the mitral valve using RT3D TEE was proposed by Salcedo et al. [19]. In principle, definition of standard views is more difficult in 3D datasets because orientation in space is unlimited compared to transthoracic 2D echo-

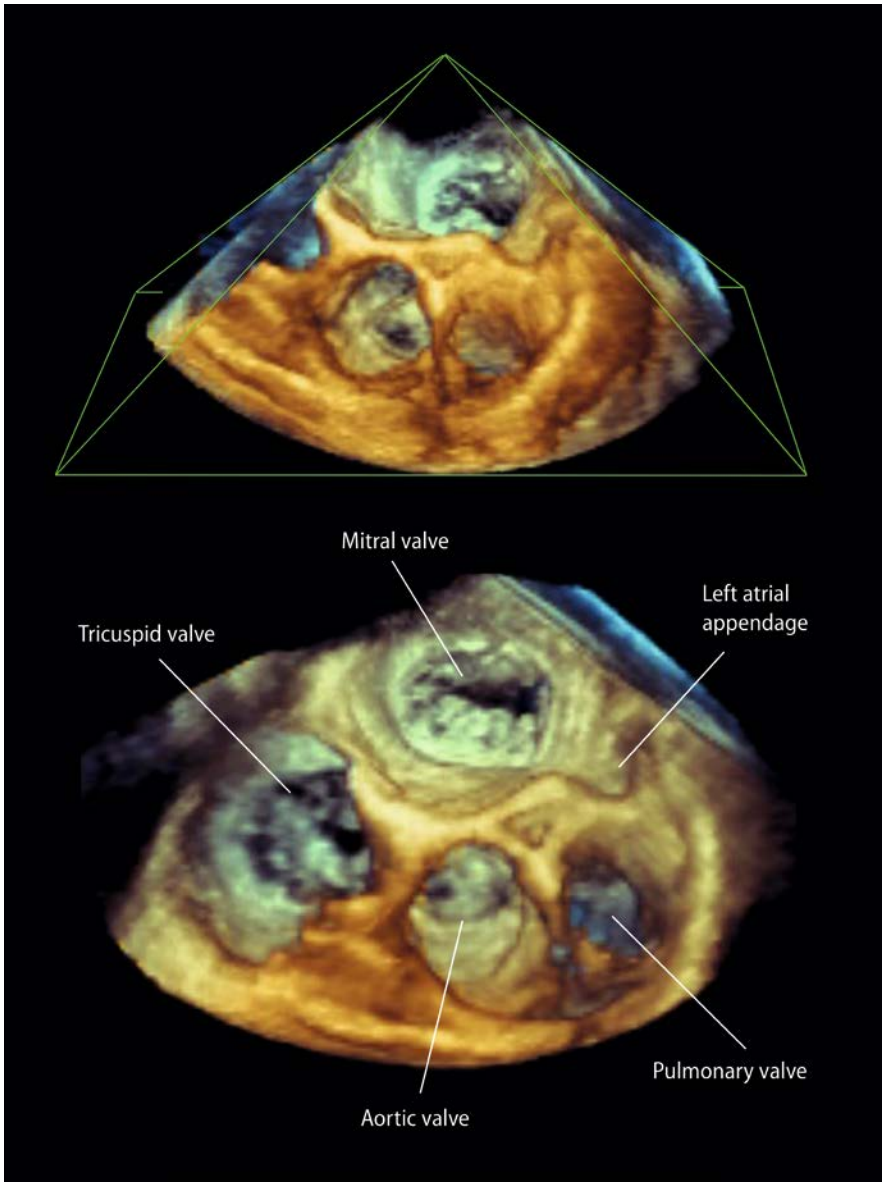

Fig. 2.22 Example of a wide transesophageal multibeat full volume dataset with good spatial and temporal resolution (25 Hz). The upper image illustrates the pyramid-shaped dataset with the left atrium near the peak of the pyramid which is from where the transducer transmits the ultrasound beam through the heart. The lower image shows the dataset in an en face view to the valve plane [→Video 2.22A,B]

cardiography which is limited to parasternal and apical views, for example.

Based on the different characteristics of the 3D acquisition modes used in transthoracic echocardiography (TTE) and transesophageal echocardiographic (TEE) examinations, the following 3D views have been found to be particularly important and useful in the majority of 3D examinations. However, it is important to emphasize that »standard 3D views« are useful, on the one hand, to follow in a structured examination protocol, but, on the other hand, it is equally important to freely rotate and orient the 3D datasets to obtain the 3D view that best describes the relevant structure or region. It is important to again emphasize that a complete 3D echocardiographic examination should consist not only of 3D en face views, but should also contain cross-sectional 2D views of structures, like mitral and aortic valves, the left atrial appendage, and the interatrial septum using either standard 2D views from conventional 2D scanning or 2D planes from a multiple planar reconstruction (MPR) mode from 3D datasets as widely used in the following chapters on special pathologies. Note that 3D image acquisition and standard 3D views as proposed in the following text do for the most part, but do not fully comply with the recent recommendations on 3D image acquisition and display [15] as for the proposed standard views we did not focus on the transducer position for volume acquisition (e.g., apical, parasternal, mid-esophageal) but more on the desired 3D image and its orientation, which was acquired from any position that provided the best 3D dataset quality.

A complete 3D echocardiographic examination should also always contain cross-sectional 2D views of structures, as 2D views add to the understanding of 3D views the same as 3D views add to the understanding of 2D views.

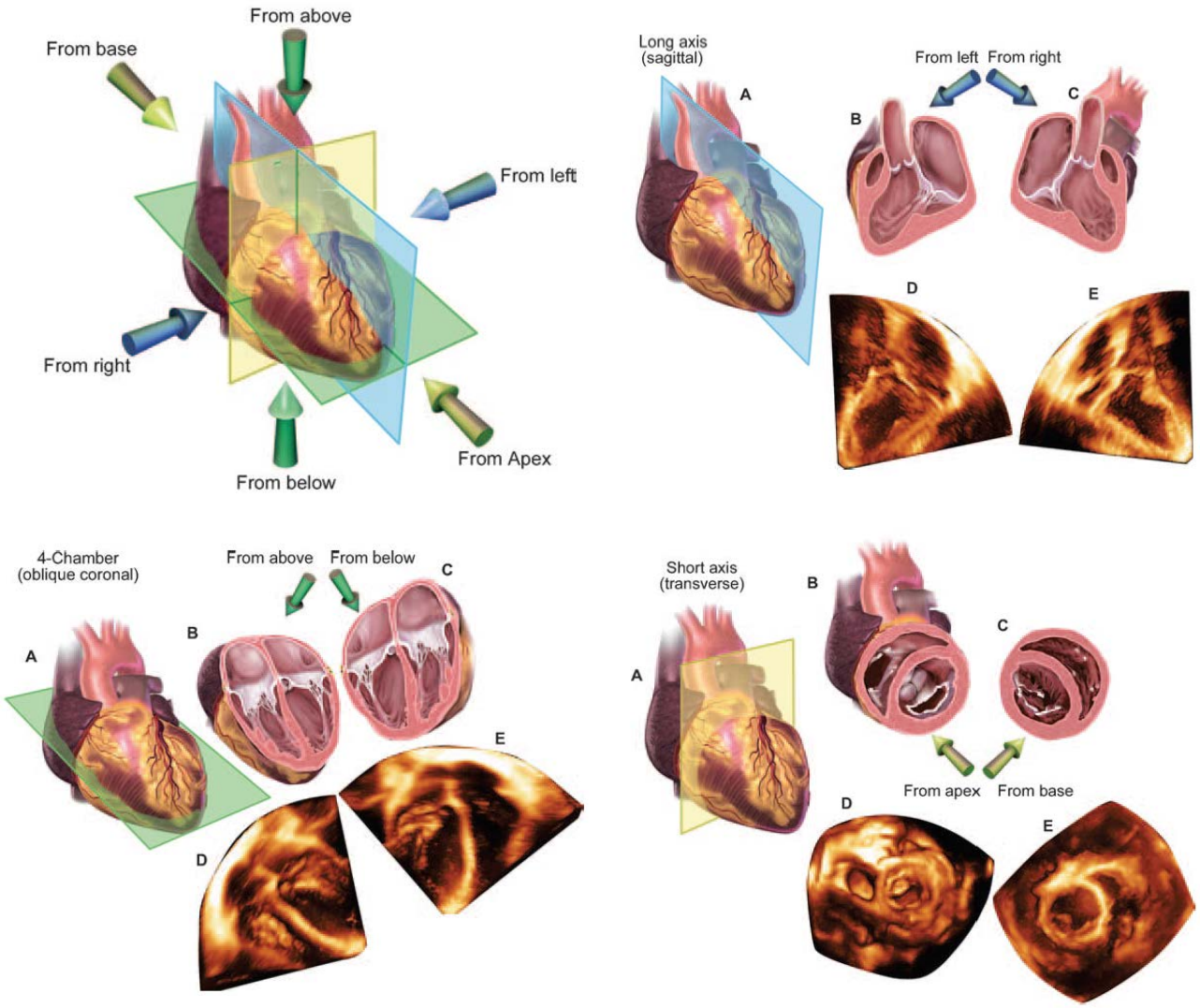

Fig. 2.23 Definition of transthoracic standard 3D views based on direct comparison of standard anatomic cut planes of the heart and corresponding 3D echocardiographic views. Reproduced from [18]; Copyright John Wiley and Sons

Standard 3D views for TTE (Fig. 2.24):

- Four-chamber view using cropped full volume (Fig. 2.24.1) provides a wide view of all four chambers as well as the mitral and tricuspid valves with good image resolution and good temporal resolution (here 20 Hz).
- Two-chamber view using cropped full volume (Fig. 2.24.2) provides a 2-chamber view of the LV and LA with good image resolution and good temporal resolution (here 20 Hz). Same dataset as used in Fig. 2.24.1. Provides en face view of interventricular septum for detection of VSD.
- En face mitral and tricuspid valve views from the apex using cropped full volume (Fig. 2.24.3) provides en face view of mitral and tricuspid valve from the LV apex with good image resolution and good temporal resolution (here 20 Hz). Same dataset as used in Fig. 2.24.1.
- En face mitral valve view from the LV using live 3D zoom (Fig. 2.24.4) provides a focused live view of the mitral valve from the LV with moderate image resolution and limited temporal resolution (here 16 Hz). Capturing part of aortic valve helps with orientation. This view is particularly useful for assessment of incomplete mitral leaflet closure and restricted mitral leaflet motion in mitral stenosis.
- En face mitral valve view from the LA using live 3D zoom (Fig. 2.24.5) provides a focused live view of the mitral valve from the LA with moderate image resolution and limited temporal resolution (here 16 Hz). Same dataset as used in Fig. 2.24.4. This view is particularly useful for assessment of organic mitral valve disease, e.g., prolapse and flail leaflet. It also provides a focused live view of the aortic valve from the aortic root. Morphologic assessment of the aortic valve, however, can be limited with current transthoracic matrix array transducer performance.
- En face aortic valve view from the LV using live 3D zoom (Fig. 2.24.6) provides a focused live view of the aortic valve and the LVOT from the LV with limited image resolution and limited temporal resolution (here 16 Hz). Same dataset as used in Fig. 2.24.4.

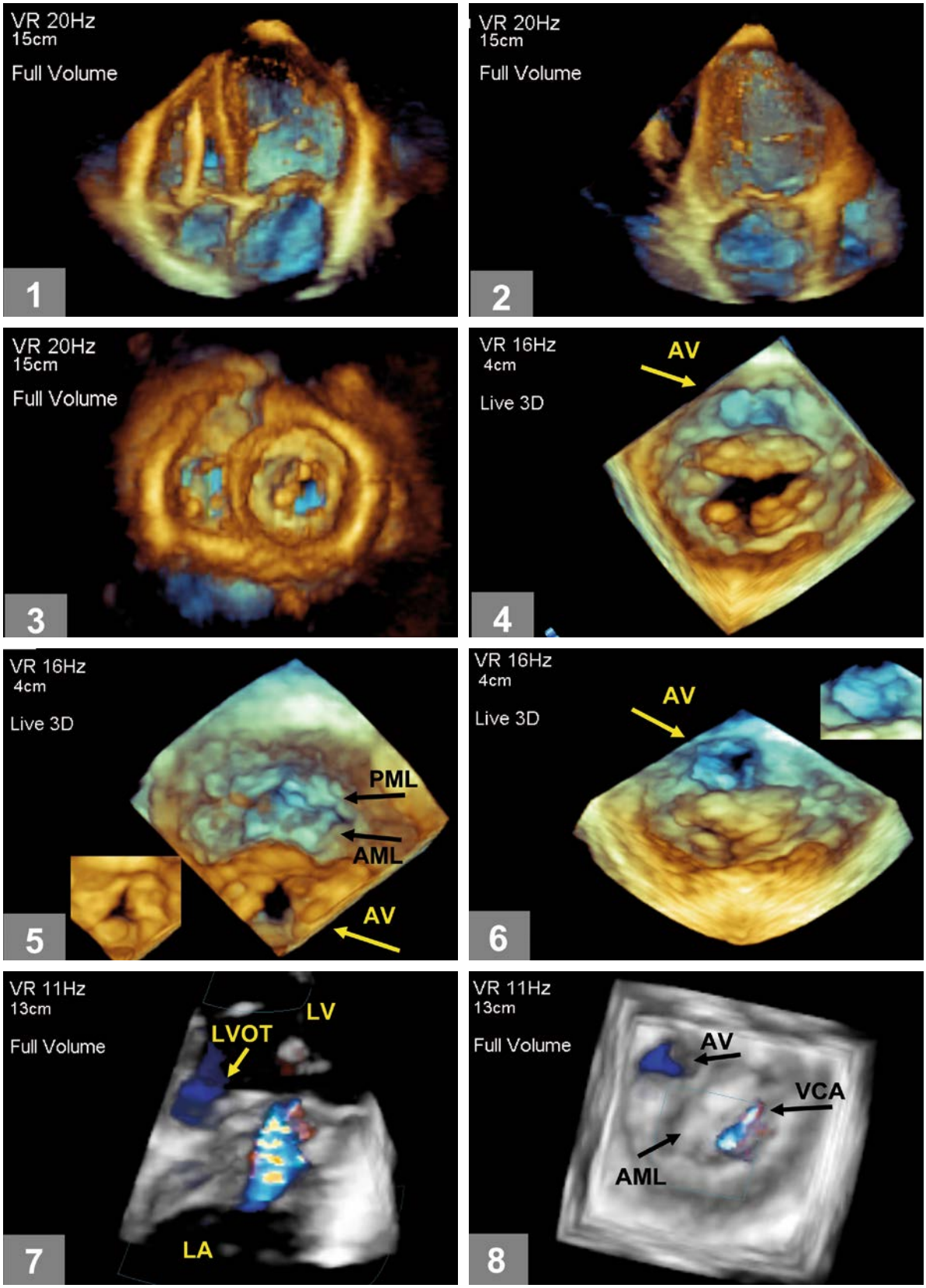

Fig. 2.24 Examples of standard 3D views for TTE according to the list given in the text. In **5** and **6**, the small pictures provide an en face view of the aortic valve in a more closed position with good depiction of the three cusps. *AV* aortic valve, *PML* posterior mitral leaflet, *AML* anterior mitral leaflet, *LVOT* left ventricular outflow tract, *LV* left ventricle, *LA* left atrium, *VCA* vena contracta area. [→Videos 2.24A–H]

- Long-axis mitral valve view using cropped full volume with color Doppler for 3D flow analysis (Fig. 2.24.7) provides 3D representation of mitral valve regurgitant and stenotic flow with good image resolution but limited temporal resolution (here 11 Hz).
- En face mitral valve view using cropped full volume with color Doppler for 3D flow analysis (Fig. 2.24.8) provides direct assessment of shape and size of vena contracta area of mitral regurgitant jet with good image resolution but limited temporal resolution (here 11 Hz). Same dataset as used in Fig. 2.24.7.

Because of currently limited 3D TTE image quality, no 3D standard views can be recommended for the assessment of narrow or thin structures like the tricuspid valve, interatrial septum, and left atrial appendage. However, this will likely change as 3D TTE image quality improves.

Standard 3D views for TEE (Fig. 2.25):

- En face mitral valve view from the LA using live 3D zoom (Fig. 2.25.1) provides a focused live view of the mitral valve with good image resolution but limited temporal resolution (here 12 Hz). Optimal narrowing of the 3D volume to the mitral valve is recommended for an acceptable volume rate. Capturing part of the AV helps with orientation. This view is particularly useful to assess organic mitral valve disease, e.g., prolapse, flail leaflet, endocarditis. It is also useful for periinterventional live monitoring and guiding during paravalvular defect closure device implantation or mitral clipping.
- En face mitral valve view from the LV using live 3D zoom (Fig. 2.25.2) provides a focused live view of the mitral valve from the LV with good image resolution but limited temporal resolution. Same dataset as used in Fig. 2.25.1. This view is particularly useful for assessment of incomplete mitral leaflet closure and restricted mitral leaflet motion in mitral stenosis.
- En face mitral valve view from the LA using full volume (Fig. 2.25.3) provides an increased view of the mitral valve, left atrial appendage, aortic valve, part of tricuspid valve, and interatrial septum from the LV with high image resolution and increased temporal resolution (here 27 Hz). Can be affected by stitching artifacts.
- En face mitral valve view using full volume with color Doppler for 3D flow analysis (Fig. 2.25.4) provides 3D representation of mitral regurgitation color Doppler jets in a relatively narrow view of the mitral valve with good image resolution but limited temporal resolution (here 14 Hz). Can be affected by stitching artifacts.
- En face aortic valve view from the aorta using a narrow live 3D mode (Fig. 2.25.5) provides a live view from the aortic root to the aortic valve with good temporal resolution (here 30 Hz). Because of the upright position, the aortic valve fits into the narrow live 3D volume.
- Long-axis aortic valve view using a narrow live 3D mode (Fig. 2.25.6) provides a long-axis live view of the aortic valve, LVOT, and aortic root with good image resolution and good temporal resolution (here 30 Hz). This live view is particularly useful for periinterventional live monitoring during transcatheter aortic valve implantation.
- En face aortic valve view from the aorta using full volume (Fig. 2.25.7) provides an en face view from the aortic root to the aortic valve with high image resolution and good temporal resolution (here 27 Hz). Can be affected by stitching artifacts. Because the dataset is not live, it cannot be used for periinterventional live monitoring.
- Long-axis aortic valve view using cropped full volume (Fig. 2.25.8) provides a long-axis view of the aortic valve, LVOT, and aortic root with high image resolution and good temporal resolution (here 27 Hz). Same dataset as used in Fig. 2.25.7 cropped to the aortic valve (can be affected by stitch artifacts and cannot be used for periinterventional live monitoring).
- Long-axis aortic valve view using cropped full volume with color Doppler for 3D flow analysis (Fig. 2.25.9) provides 3D representation of aortic valve regurgitant and stenotic flow with good image resolution but limited temporal resolution (here 14 Hz). Can be affected by stitch artifacts.
- Left atrial appendage view from the LA using the live 3D zoom (Fig. 2.25.10) provides a focused live view from the LA to left atrial appendage with good image resolution and moderate temporal resolution (here 15 Hz). Capturing part of the mitral valve (MV) helps with orientation. This view is particularly useful for periinterventional live monitoring during left atrial appendage closure device implantation.
- Atrial view to the interatrial septum using live 3D zoom (Fig. 2.25.11) provides a live view of the RA, LA, and interatrial septum with good image resolution but limited temporal resolution (here 11 Hz). This view is particularly useful for periinterventional live monitoring during interatrial septum closure device implantation, guiding transseptal puncture or guiding of EP catheter positioning.
- Interatrial septum view with contrast application for patent foramen ovale (PFO) testing using live 3D zoom (Fig. 2.25.12) provides a live view from the LA to the interatrial septum. Similar datasets, such as in Fig. 2.25.11, could also be used. This view is particularly useful for detection and localization of transseptal contrast bubble passage through a PFO. This view is also useful for periinterventional live monitoring during interatrial septum closure device implantation and transseptal puncture.

The order of the views and acquisition of datasets are interchangeable according to the leading pathology.

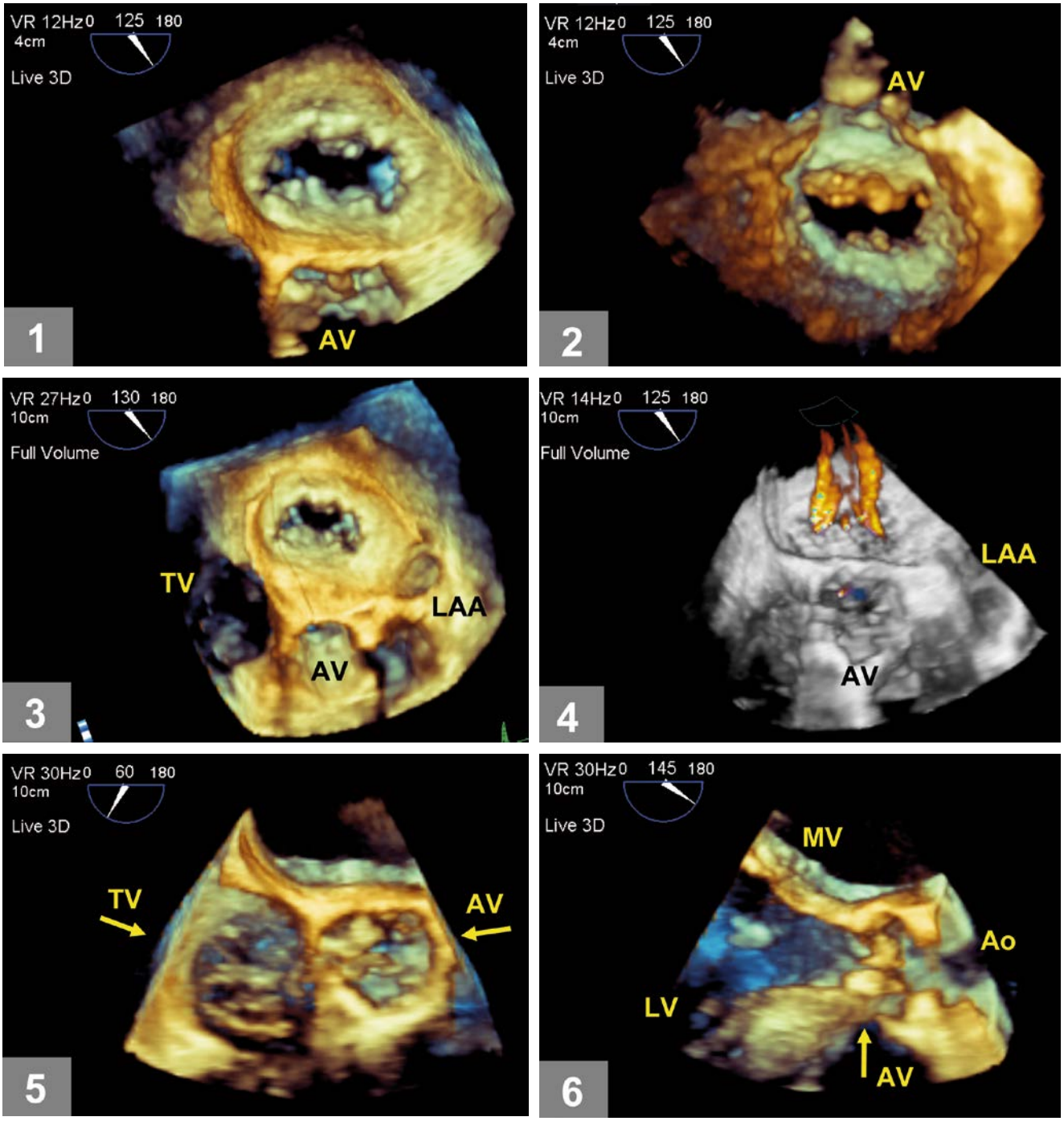

Fig. 2.25 Examples of standard 3D views for TEE according to the list given in the text. *AV* aortic valve, *TV* tricuspid valve, *LAA* left atrial appendage, *MV* mitral valve, *LV* left ventricle, *Ao* aorta, *AML* anterior mitral leaflet, *LA* left atrium, *IAS* interatrial septum, *RA* right atrium, *SVC* superior vena cava. [→Videos 2.25A–L]

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