

Three-Dimensional Echocardiographic Technology

Ivan S. Salgo, MD, MS

*Cardiovascular Investigations, Ultrasound Research & Development, Philips Medical Systems,
3000 Minuteman Road, Andover, MA 01810, USA*

Today echocardiography remains the most commonly requested imaging study of the heart. Moreover, recent insights into the understanding of the sophistication of the structural and functional mechanisms of the heart have come from advances in this area. Combined with portability, safety, low cost, and widespread availability, it is a tribute to the evolution of ultrasound technology that echocardiography's role in deciphering clinical problems in cardiac physiology has kept pace with the ever-increasing demands of a busy practice. This article addresses the current state of the art of technology in 3D echocardiography as it applies to transducer design, beam forming, display, and quantification. Because 3D echocardiography encompasses many technical and clinical areas, this article reviews its strengths and limitations and concludes with an analysis of what to use when. In this article, the author will be precise about the terminology related to 3D imaging and use dimension to refer to one, two, or three spatial dimensions.

Transducer design

The ultrasound transducer keeps echocardiography unique among its siblings in imaging technology. It converts electrical energy into mechanical vibrations and vice versa. To understand what sets 3D systems apart from conventional scanning systems, one needs to review some acoustic principles. Current 2D systems transmit and receive acoustic beams in a flat 2D scanning plane. As opposed to M mode (one spatial and

one temporal dimension), 2D scanning systems sweep a scan line to and fro within this 2D imaging plane; the angular position of the beam is said to vary in the azimuthal dimension. Even though traditional, flat 2D scanning comprises two spatial dimensions plus one temporal dimension, this is not 3D imaging. The transducer itself consists of elements working in concert to create a scan line. Typically, a conventional transducer consists of 64 to 128 elements spaced according to the ultimate frequency (and hence wavelength) of the acoustic vibrations; these propagate radially along the direction of the scan line. This array of elements steers the outward ultrasound beam or scan line by using interference patterns generated by varying the spatiotemporal phase of each element's transmit event. These principles comprise the underpinnings of any phased array system.

Armed with this knowledge, one can survey the key differences pertinent to 3D ultrasound imaging. First, these 64 to 128 elements of a conventional cardiac transducer are historically arranged along a single row. Technically this is referred to as a one-dimensional array of elements. (Remember, the two spatial dimensions in the image come from sweeping the beam by firing along this row at different times.) Nonetheless, the world has three spatial dimensions. Note that this flat scanning plane of ultrasound energy is not perfectly flat in three dimensions, even though the image appears so on the scanning system. An innovation applied in the last decade was to take just five to seven rows of elements to keep the beam flatter. These few stacks of rows now comprise for example a 5×64 or 320 element or primitive matrix array transducer. These rows, however, were not electrically independent from each other. These primitive matrix array transducers thinned the beam by growing the

This work was supported in part by NIH grant RO1 HL073647.

E-mail address: ivan.salgo@philips.com

aperture in the elevation dimension. They did not however, steer in the third dimension.

True 3D ultrasound steering has been the subject of much academic and industrial research that began in the 1980s. To actually steer an ultrasound beam, a 2D matrix array comprising as many elements in the elevational dimension as in the azimuthal dimension needed to be fabricated. To do this, a block of transducer material was diced by a diamond tipped saw to create each element (Fig. 1). Today's transducers comprise more than 3000 elements and can have row and column sizes greater than 60. Note that this is a 2D matrix array that generates 3D images. This concept was known for decades. The significant innovation that actually allows steering is making the elements electrically independent from each other. This allows generating a scan line that varies azimuthally and elevationally. Thus a true 3D scan line is born. Early transducers, while having electrically independent rows, did not have every element electrically active, as the technology to connect such a dense array was not known yet. Newer types of electrical circuitry connecting each element first were commercialized in 2002. Today, miniaturization has allowed fitting thousands of fully sampled elements into the tip of a transesophageal transducer (Fig. 2). It is important to realize that the physical aperture needs to be designed according to the application. The wider the aperture in each dimension, the better the scan line can be focused. Transthoracic and transesophageal imaging, however, is physically limited by the

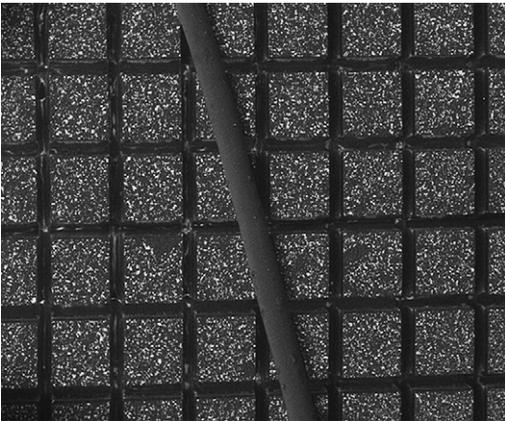


Fig. 1. Scanning electron micrograph of a fully sampled 2D matrix array used for 3D beam steering. A human hair is shown within the view for comparison.



Fig. 2. A Matrix Array Transesophageal Transducer distal tip. The active aperture is 10 mm \times 10 mm. Note that this supports 2D and 3D imaging, and spectral and color Doppler modes.

width and length of the transducer surface, and frequency aperture tradeoffs are taken into account when the transducer is designed for each application.

Modern 2D transducers therefore consist of thousands of electrically active elements that steer a scan line left and right as well as up and down. New materials that allow more bandwidth (simultaneous high and low frequencies) allow these matrix array transducers to obtain both penetration and high-resolution imaging.

Beam forming in three spatial dimensions

Beam forming constitutes the steering and focusing of transmitted and received scan lines. Because each element must have independent electrical control by the ultrasound system, a conventional cable that would be used to connect each element would make the transducer cable unwieldy. To reduce the size of the cable and reduce power consumption, a significant portion of the beam steering is done within the transducer in highly specialized integrated circuits. The main system steers at coarse angles, but the transducer circuits steer in fine increments in a process termed microbeam forming. Summing is the act of combining raw acoustic information from each element to generate a scan line and by summing these in a sequence (first in the transducer and then subsequently in the system). Both the transducer connectors and cable are reduced in size dramatically. The 3D beam former steers both in azimuth and elevation. This creates a 3D spherical

wedge of acoustic information that subsequently is processed.

The radiofrequency (RF) data are summed and processed using various signal techniques and finally put into rectangular (Cartesian) space by a 3D scan converter. Note that this is an extension of the traditional phased array approach. Older 3D techniques used in the 1990s did not steer electronically but rather involved combining 45 to 100 cardiac loops and recombining them to create a 3D data set [1]. Although this works for static structures, irregular rhythms spell the downfall of gated reconstruction of many beats. Any transducer movement during acquisition of these many gates created artifacts and required a significant amount of smoothing of the data. Although this helped reduce attention to slippage or misalignment, it reduced the overall image quality if significant stitching needed to be done.

There are two major black and white modes run in an electronically steered 3D system. The first is a live mode where the system scans in real-time three dimensions. The test of this mode is as follows: if the transducer comes off the chest, the image disappears. As with sector scanning, the volume pyramid may be reduced to zoom in three dimensions. This is again a live mode. Today computer and beam forming processing power are not limiting for 3D ultrasound image generation; it is the speed of sound. Gating, this time only four to eight beats, allows a technique to generate wider volumes while maintaining frame rate. It is done by stitching four (or more) gates together in full-volume mode. This can generate greater than 90-degree scanning volumes at frame rates greater than 30 Hz. In patients who have arrhythmias, respiratory rate (RR) intervals that fall out of a set range cause the discard of the errant sub-volume and the system scans again until all sub-volumes are generated from suitable beat intervals. Thus in patients who have irregular intervals, as long as the average RR falls within a reasonable range, a full volume can be reconstructed. Because Doppler methods require many transmit events to be fired along a single path, color Doppler methods use gating. These multiple events are compared or cross-correlated to find the velocity of moving structures. The true 3D spatial modes may be summarized as follows (Figs. 3–5):

- Live 3D mode—instantaneous
- 3D zoom—instantaneous
- Full-volume—gated
- 3D color Doppler—gated

A variant of pyramidal 3D beam forming is to scan simultaneous planes in a process known as bi- and triplane imaging. Because this process requires firing fewer scan lines, frame rate is increased at the cost of spatial resolution. Here is a point worth emphasizing: all 3D echocardiography is subject to the laws of physics. Artifacts such as ringing, reverberations, shadowing, and attenuation occur in three and two dimensions. The system must play within these constraints. The numbers and densities of scan lines can be traded off to increase temporal resolution. The constraints of a 3D image are bounded by: (1) frame rate, (2) 3D volume size, and (3) image resolution. Increasing the requirement of one of these causes a drop in the other two, all things being equal. Systems can generate synthetic receive lines (four times, 16 times, or 32 times) for each transmit line by increasing beam forming parallelism. Note that as the interpolated receive line is steered farther and farther away from the transmitted scan line, the signal-to-noise ratio drops. Thus there is a practical limit to receive parallelism. The benefit is that it can increase the density of received lines, thereby improving resolution.

Display of 3D information

2D computer displays consist of rows and blocks of pixels (picture elements) that comprise a 2D image. A 3D data set consists of bricks of pixels called volume elements or voxels. Because the computer screen is only 2D in nature, perspective is used to simulate the appearance of 3D depth, as an object is virtually deeper or farther away from the viewing screen. Essentially the 3D data set (ie, collection of voxels) can be rotated with respect to the computer screen. Moreover, a process known as cropping can be used to cut into the volume and make some voxels invisible; for example, one can cut away the left atrium to see the mitral valve. After rotation and cropping, the remaining voxels are ready to be projected onto pixels in the same way as 3D objects in a room are projected onto the retina in two dimensions.

3D data sets of voxels are turned into 2D images in a process known as volume rendering. The 2D image appears in three dimensions because of perspective. There are several different algorithms to accomplish this conversion (eg, ray casting, shear-warp, and others), but they

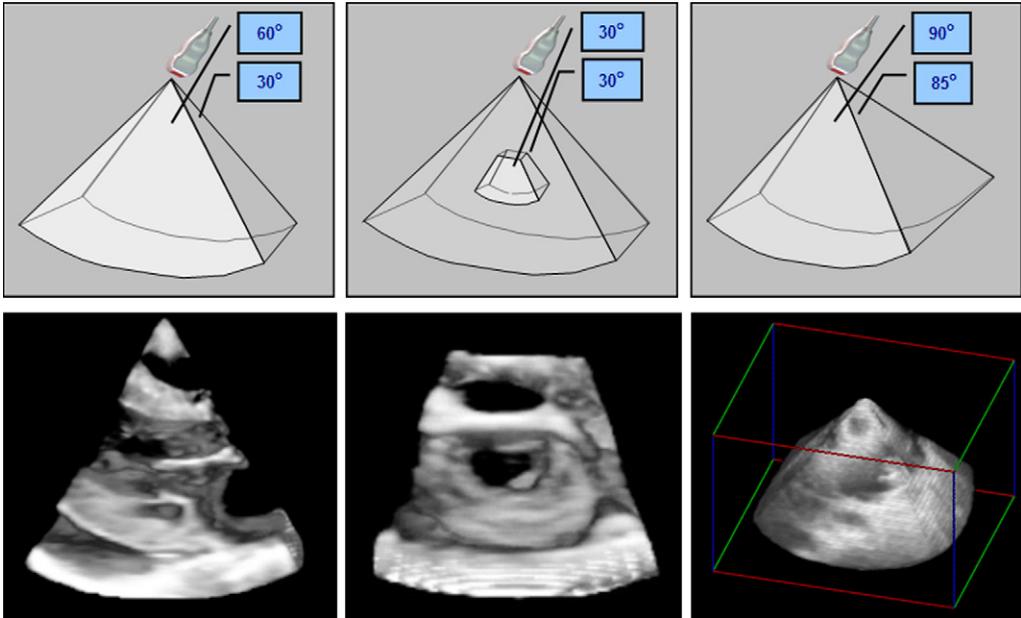


Fig. 3. 3D scanning volumes shown as volume rendering. These show the extent of the 2D scan line steering limits in each case. Ultimately, these will constitute a voxel data set. (Courtesy of the University of Chicago Medical Center, Chicago, IL; with permission.)

essentially cast a light beam through the collection of voxels. Either the light beam hits enough tissue so as to render it opaque (eg, tissue), or it keeps shining through transparent voxels so as to render it transparent (eg, blood pool). Tissue surfaces derived from voxel data sets sometimes are termed surface rendering, but this is inaccurate terminology. Additional algorithms can be used to increase the perception of depth by applying different

hues on the front (near the screen) of the data set as opposed to voxels far from the screen.

Quantification in three dimensions

Although visualization of anatomy in its true 3D state is important, many physicians believe that the most significant value 3D echocardiography

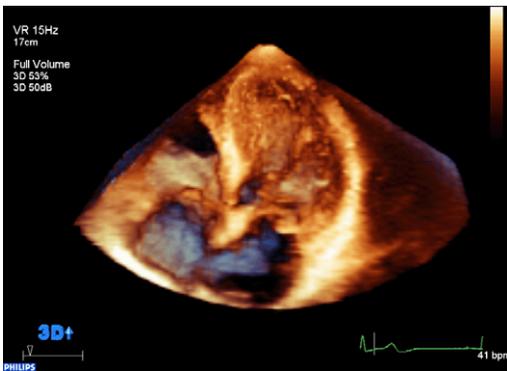


Fig. 4. Full volume acquisition encompassing the left ventricle. Depth-dependent dynamic colorization has been used to code hue according to depth perspective from the viewer. This adds visual cues to increase the “3D sense.”

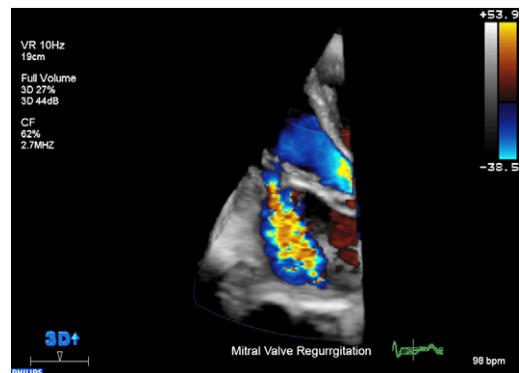


Fig. 5. Gated 3D color acquisition. A 3D jet of mitral regurgitation is shown. The classic 2D color map has been adapted to render color voxels showing the 3D nature of the jet.

has for adult echocardiography is quantification. True myocardial motion occurs in three dimensions, and traditional 2D scanning planes do not capture the entire motion of the heart or else move or slip while scanning. Quantifying implies segmenting structures of interest from the 3D voxel set. Although the voxels themselves can be tagged, for example coloring the right ventricular (RV) voxels separately from the left ventricular (LV) voxels, computer vision techniques frequently employ methods that define an interface (eg, the LV endocardial border). This interface typically is constructed as a mesh of points and lines and displayed in a process known as surface rendering (Fig. 6). Automobile, building, and engineering parts are rendered as meshes by surface rendering in computer-aided design tools.

3D quantification of the left ventricle typically employs a surface-rendered mesh. This allows accurate computation of volume, regional wall motion, and regional synchrony. Because the entire extent of the left ventricle is taken into account, no foreshortening errors or assumption of LV volume are generated [1,2]. Technically, a 3D deformable model is used to find the LV endocardial surface in three dimensions. This is the most accurate way to quantify LV volumes. Moreover, 3D LV remodeling can be displayed parametrically using differential geometry techniques (Fig. 7). Bi- and triplane methods help avoid foreshortening errors, but if an aneurysmal dilatation occurs between planes, the computed LV volumes will have some interpolation error. Quantification of LV synchrony is possible in

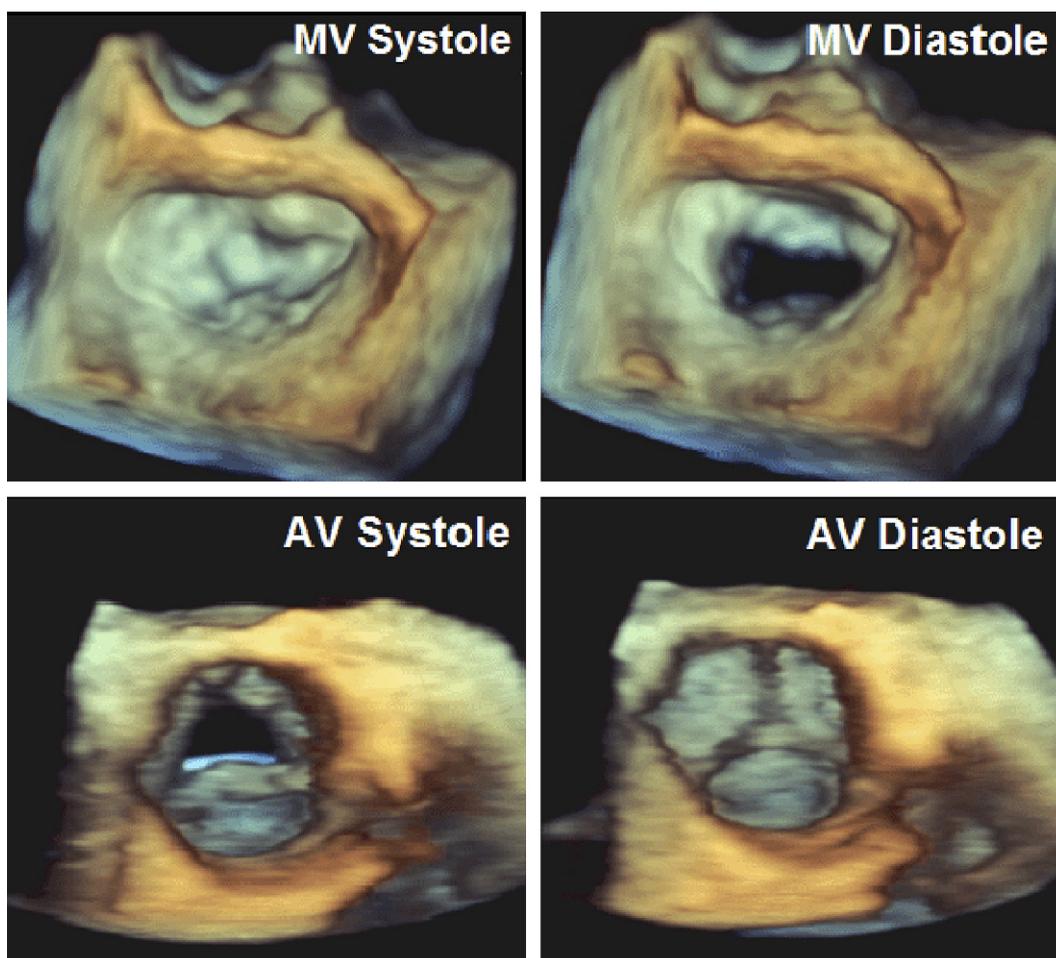


Fig. 6. 3D matrix TEE acquisitions of the aortic and mitral valves. (Courtesy of the University of Chicago Medical Center, Chicago, IL; with permission.)



Fig. 7. 3D semi-automated border detection using surface rendering of the left ventricular shell to display its 3D nature in QLAB. Each segment has been analyzed for regional ejection, and segmental function is shown by the waveforms.

three dimensions also [3]. The required frame rate depends on the questions being asked. A frame rate of 30 Hz (33 milliseconds between frames) is inadequate to quantify intramyocardial motion; this is better suited to be studied by tissue Doppler or speckle tracking techniques. Regional synchrony, however, can be measured by three dimensions, because it assesses blood ejection not tissue motion. Because the ejection curve is naturally smooth, it requires less frame rate (has a lower band limit). Moreover, the Nyquist theorem frequently is misrepresented in this area. Some investigators argue that 20 millisecond phenomenon (eg, regional phase differences in peak ejection at different segments) cannot be quantified at 33 millisecond frame intervals. This is an incomplete statement. In fact, if the waveform's upper natural frequency limit reaches its maximum at 15 Hz (half the sampling frame rate) the waveform is fully determined. Thus increasing the frame rate to 100 Hz (10 millisecond intervals) would yield no additional information if the wave already was sampled fully at 30 Hz. This appears to be the case for many dyssynchrony

patients and is why 3D echo stratifies dyssynchrony successfully.

New 3D electronically steered transesophageal transducers are yielding ultrasound images never before seen on the beating mitral valve (Fig. 8). This also allows the mitral apparatus to be segmented at end-systole with great accuracy. The true three dimensional nature of the mitral annulus, leaflets, and chordal apparatus can be measured. This further allows sophisticated analyses of the nonplanar shape of the mitral annulus [4,5]. These 3D measurements include: annular diameters, annular nonplanarity, commissural lengths, leaflet surface areas, aortic to mitral annular orientation.

Other areas subject to 3D quantification include quantifying volumes of all chambers of the heart (ie, the left and right atria and right ventricle). Quantification of 3D color flow is an active area of research [6]. One of the limitations for color flow quantification includes measuring unsteady flow (as in an intermittent jet of mitral regurgitation) that may not be sampled adequately by gated 3D color Doppler methods.

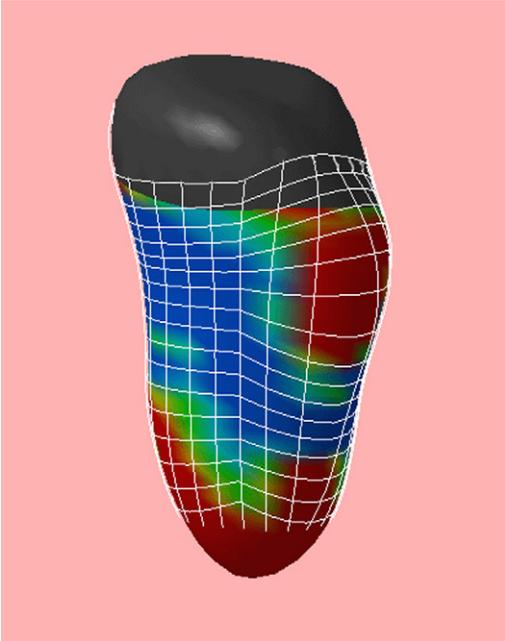


Fig. 8. 3D surface-rendered mesh from QLAB has been analyzed using principal curvature analysis. An aneurysmal portion of the anterior basal segment is shown in red. Note that the apex is normally red because of its high degree of curvature. Additionally, a set of white grid lines applying geodesic computation shows the bulge at the aneurysmal segment.

What to use when

As 3D technology increases in sophistication, so do the new physiologic parameters that can be measured [7]. It is important to begin with anatomic or physiologic questions and turn to 3D as needed to answer these questions with accuracy. The single most requested measurement for any indication for an echo examination is assessment of LV function and size. This should be measured by three dimensions, and techniques to assess LV endocardial volume, mass and regional function are readily available [8–10]. 3D techniques are more accurate than multiplane techniques, which more accurate than single plane methods. Echo contrast can be used to increase volume accuracy in patients who have poor acoustic windows [11]. Stress echo examinations can benefit from 3D or triplane methods [12,13]. This is because single plane scanning does not assure with quantitative certainty that the true apex of the heart is within the scanning volume. At this point, the application of 3D technology depends on the referring question.

Patients who undergo mitral valve repair should have intraoperative transesophageal echo as part of their care. 3D methods now allow leaflet anatomy to be displayed functionally with resolution that was not possible before. Although conventional 2D echocardiography is likely sufficient for the detection of thrombus and vegetations, 3D echocardiography allows visualization and quantification of the mitral apparatus and tricuspid apparatus in their living 3D state [14,15]. Live 3D transesophageal echo (TEE) is especially useful in delineating areas of leaflet prolapse as in patients who have Barlow's disease. Moreover, 3D echo is useful in assessing the degree of leaflet restriction within the apparatus and annular changes (Fig. 9) [16,17]. Sophisticated changes in LV remodeling can be assessed in ischemic mitral regurgitation [18,19]. Changes of mitral leaflet systolic anterior motion can be assessed in hypertrophic cardiomyopathy [20].

3D visualization is especially useful in identifying intracardiac problems such as atrial septal defects. This is especially useful in identifying and characterizing congenital defects for children and the growing population of adults with congenital heart disease.

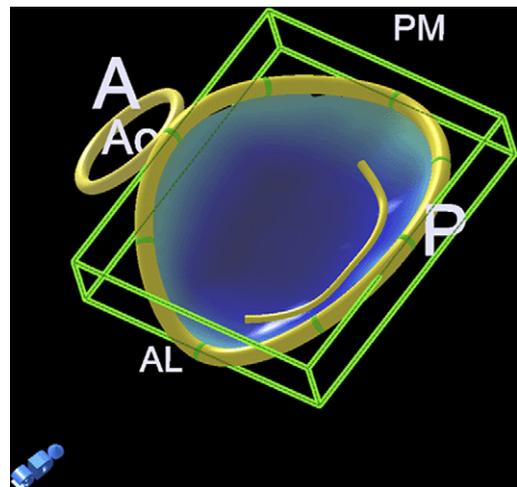


Fig. 9. 3D mitral valve quantification from matrix TEE images (QLAB). The mitral annulus in relation to the aortic annulus is shown. Additionally, coaptation from commissure to commissure is shown. Leaflet segmentation allows computation of leaflet restriction and prolapse area. The green box displays the nonplanarity of the mitral annulus in 3D space.

The future for 3D echocardiography looks bright. Advances in technology will allow larger scanning volumes and more sophisticated methods of quantification, such as 3D speckle tracking and analysis of LV torsion and mechanics. One of the most exciting areas includes the use of 3D echocardiography to guide intracardiac procedures without the need for cardiopulmonary bypass [21,22]. Placement of atrial septal defect (ASD) devices and percutaneous valve therapies likely will benefit from the live nature of 3D TEE imaging and broaden not only the diagnostic uses of 3D echocardiography but those used for therapy.

Acknowledgment

The author gratefully acknowledges the support and friendship of Bernard Savord and Karl Thiele in preparation of this manuscript.

References

- [1] Salgo IS. Three-dimensional echocardiography. *J Cardiothorac Vasc Anesth* 1997;11(4):506–16.
- [2] Gerard O, Billon AC, Rouet JM, et al. Efficient model-based quantification of left ventricular function in 3D echocardiography. *IEEE Trans Med Imaging* 2002;21(9):1059–68.
- [3] Kapetanakis S, Kearney MT, Siva A, et al. Real-time three-dimensional echocardiography: a novel technique to quantify global left ventricular mechanical dyssynchrony. *Circulation* 2005;112(7):992–1000.
- [4] Salgo IS, Gorman JH III, Gorman RC, et al. Effect of annular shape on leaflet curvature in reducing mitral leaflet stress. *Circulation* 2002;106(6):711–7.
- [5] Watanabe N, Ogasawara Y, Yamaura Y, et al. Mitral annulus flattens in ischemic mitral regurgitation: geometric differences between inferior and anterior myocardial infarction: a real-time 3-dimensional echocardiographic study. *Circulation* 2005;112(Suppl 9):1458–62.
- [6] Pemberton J, Ge S, Thiele K, et al. Real-time three-dimensional color Doppler echocardiography overcomes the inaccuracies of spectral Doppler for stroke volume calculation. *J Am Soc Echocardiogr* 2006;19(11):1403–10.
- [7] Lang RM, Mor-Avi V, Sugeng L, et al. Three-dimensional echocardiography: the benefits of the additional dimension. *J Am Coll Cardiol* 2006;48(10):2053–69.
- [8] Thomas JD, Popovi ZB, Thomas JD, et al. Assessment of left ventricular function by cardiac ultrasound. *J Am Coll Cardiol* 2006;48(10):2012–25.
- [9] Mor-Avi V, Sugeng L, Weinert L, et al. Fast measurement of left ventricular mass with real-time three-dimensional echocardiography: comparison with magnetic resonance imaging. *Circulation* 2004;110(13):1814–8.
- [10] Sugeng L, Mor-Avi V, Weinert L, et al. Quantitative assessment of left ventricular size and function: side-by-side comparison of real-time three-dimensional echocardiography and computed tomography with magnetic resonance reference. *Circulation* 2006;114(7):654–61.
- [11] Corsi C, Coon P, Goonewardena S, et al. Quantification of regional left ventricular wall motion from real-time 3-dimensional echocardiography in patients with poor acoustic windows: effects of contrast enhancement tested against cardiac magnetic resonance. *J Am Soc Echocardiogr* 2006;19(7):886–93.
- [12] Malm S, Frigstad S, Sagberg E, et al. Real-time simultaneous triplane contrast echocardiography gives rapid, accurate, and reproducible assessment of left ventricular volumes and ejection fraction: a comparison with magnetic resonance imaging. *J Am Soc Echocardiogr* 2006;19(12):1494–501.
- [13] Yang HS, Pellikka PA, McCully RB, et al. Role of biplane and biplane echocardiographically guided 3-dimensional echocardiography during dobutamine stress echocardiography. *J Am Soc Echocardiogr* 2006;19(9):1136–43.
- [14] Garcia-Orta R, Moreno E, Vidal M, et al. Three-dimensional versus two-dimensional transesophageal echocardiography in mitral valve repair. *J Am Soc Echocardiogr* 2007;20(1):4–12.
- [15] Ton-Nu TT, Levine RA, Handschumacher MD, et al. Geometric determinants of functional tricuspid regurgitation: insights from 3-dimensional echocardiography. *Circulation* 2006;114(2):143–9.
- [16] Watanabe N, Ogasawara Y, Yamaura Y, et al. Quantitation of mitral valve tenting in ischemic mitral regurgitation by transthoracic real-time three-dimensional echocardiography. *J Am Coll Cardiol* 2005;45(5):763–9.
- [17] Yamaura Y, Watanabe N, Ogasawara Y, et al. Geometric change of mitral valve leaflets and annulus after reconstructive surgery for ischemic mitral regurgitation: real-time 3-dimensional echocardiographic study. *J Thorac Cardiovasc Surg* 2005;130(5):1459–61.
- [18] Hung J, Guerrero JL, Handschumacher MD, et al. Reverse ventricular remodeling reduces ischemic mitral regurgitation: echo-guided device application in the beating heart. *Circulation* 2002;106(20):2594–600.
- [19] Messas E, Yosefy C, Chaput M, et al. Chordal cutting does not adversely affect left ventricle contractile function. *Circulation* 2006;114(Suppl 1):1524–8.
- [20] Song JM, Fukuda S, Lever HM, et al. Asymmetry of systolic anterior motion of the mitral valve in patients with hypertrophic obstructive cardiomyopathy: a real-time three-dimensional

- echocardiographic study. *J Am Soc Echocardiogr* 2006;19(9):1129–35.
- [21] Cannon JW, Stoll JA, Salgo IS, et al. Real-time three-dimensional ultrasound for guiding surgical tasks. *Comput Aided Surg* 2003;8(2):82–90.
- [22] Suematsu Y, Takamoto S, Kaneko Y, et al. Beating atrial septal defect closure monitored by epicardial real-time three-dimensional echocardiography without cardiopulmonary bypass. *Circulation* 2003; 107(5):785–90.