JOURNAL OF CARDIOVASCULAR DIAGNOSIS AND PROCEDURES
Volume 15, Number 2, 1998
Mary Ann Liebert, Inc.

Real-Time 3D Ultrasound: A New Look at the Heart

GEORGE D. STETTEN,[1] TAKAHIRO OTA,[2] CHIKAI J. OHAZAMA,[1] CRAIG FLEISHMAN,[2] JOHN CASTELLUCCI,[1] JOHN OXAAL,[3] THOMAS RYAN,[2] JOSEPH KISSLO,[2] and OLAF T. VON RAMM[1]

ABSTRACT

Matrix array ultrasound is a medical imaging modality in which a 3D volume is scanned electronically without physically moving the transducer, permitting rapid continuous 3D scanning of the heart. Unlike reconstructive 3D ultrasound, which relies on physically moving a linear array and acquires data during multiple cardiac cycles gated to the ECG, matrix array ultrasound has no moving parts, resulting in a scan rate rapid enough (22 frames/second) to smoothly sample heart motion within a single cardiac cycle. Therefore, these cardiac studies have been described as real time, and the modality itself has been labeled Real-Time 3D (RT3D) ultrasound. We review the first application of matrix array ultrasound to *in vivo* cardiac imaging of normal volunteers, describing methods of displaying the data during the scan, as well as afterwards on a graphics replay station. We conclude that by introducing the capability of real-time 3D cardiac imaging, matrix array ultrasound provides an important new clinical tool.

INTRODUCTION

At Duke University we have developed an imaging technology known as Real Time 3D (RT3D) ultrasound, based on a matrix array transducer that scans a 3D volume electronically.[1–5] Replacing the single row of elements found in conventional linear (1D) transducers (see Fig. 1A), the elements in a matrix array transducer are arranged in a two-dimensional grid (see Fig. 1B). As with the linear array, the direction in which the matrix array transmits and receives ultrasound energy is controlled by timing individual transducer elements during transmission and reception of the ultrasound. With a linear array, only the direction within a slice, the so-called azimuth, can be controlled, whereas a matrix array offers steering in both the beam's azimuth and elevation, permitting interrogation of an entire pyramid-shaped volume. Using the prototype of matrix-array ultrasound machine known as "T4" (see Fig. 2), we have conducted *in vivo* studies and explored methods of displaying the volumetric data.

[1]Department of Biomedical Engineering and [2]Department of Cardiology, Duke University, Durham, NC.
[3]Volumetrics Medical Imaging, Inc., Durham, NC 27701.

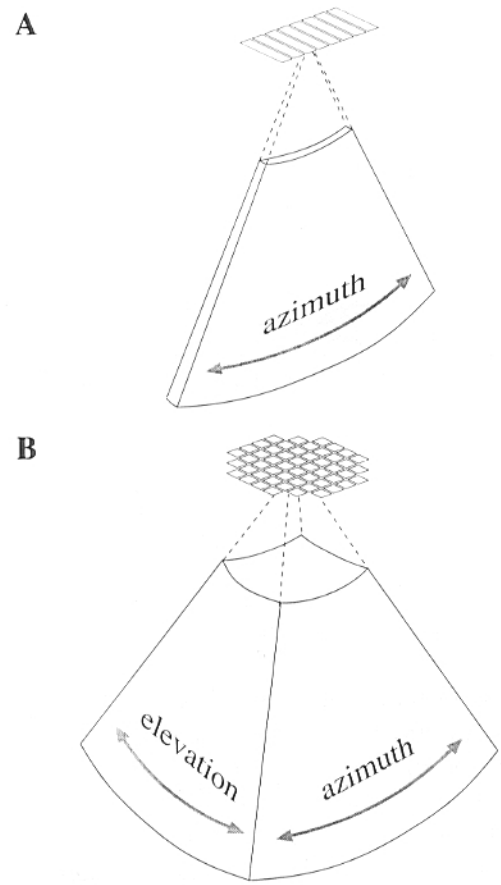

FIG. 1. (**A**) A linear (1D) array can focus within a single slice, controlling the azimuth of the beam by varying the phase between transducer elements. (**B**) A matrix array adds another dimension, permitting control of the beam's elevation and allowing the scanner to focus anywhere within the pyramid-shaped volume.

Scan versus view

In the routine description of medical images, clinicians follow standard conventions as a matter of expedience and to reduce ambiguities in their reports. The accepted terminology develops only through practice and differs with each imaging modality. Such a standard has yet to evolve for 3D ultrasound but we find it convenient to differentiate between the *scan* and the *view*. The reason for this distinction is that the anatomical content of a 3D ultrasound image is determined by two separate factors: (1) the transducer location with respect to the patient, and (2) the slice location within the data pyramid.

In this article we use *scan* to denote transducer location. As in conventional echocardiography, a parasternal scan means the transducer is aimed between the ribs adjacent to the sternum, and an apical scan means the transducer is positioned below the ribs, pointing upwards at the apex of the heart. The particular scan is established when the data is collected and cannot be altered thereafter.

We use *view* to denote which slice (or sub-volume) within the data pyramid is displayed. The view can be determined after the data are captured or by adjusting the display parameters during the scan without moving the transducer. Possible views include the basic slice orientation: B-mode, C-mode, and inclined C-mode (I-mode). Further categorization may relate the view to the underlying anatomy, such as *4-chamber, short-axis*, or *long-axis*, terms already familiar from 2D ultrasound. With volumetric displays, orientation with respect to the viewer must also be specified, for example, to differentiate between *short axis from apex and short axis toward apex.*

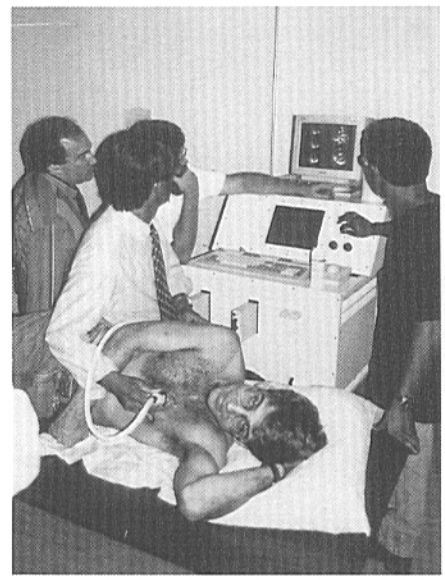

FIG. 2. A parasternal scan in progress using the T4 scanner.

MATERIALS AND METHODS

T4 scanner

The T4 ultrasound scanner contains 256 independent transmit channels and 256 independent receive channels. Transducer elements are cut in a matrix with each element being square and approximately 300 μm on a side. A typical transducer array is circular with a diameter of approximately 25 mm and an active aperture of 15 cm, operating at a frequency between 2.0 and 3.5 MHz. Because of 16:1 parallel processing,[1] T4 can accomplish a typical cardiac scan of a 13 cm deep, 64° × 64° pyramid-shaped volume containing 4,096 receive paths at a rate of 22 volumes per second. The scan rate depends on scan depth, with scan rates of 40 volumes per second possible at 6 cm, but only 14 volumes per second at 20 cm.

An important feature of scanning with a matrix array is that, unlike scanning with a conventional linear array, dynamic focusing is possible in both lateral (cross-beam) dimensions. Therefore, the beam pattern from the matrix array is circularly symmetric in the cross-beam dimensions (see Fig. 3B). Rectangular apertures in linear arrays result in beam patterns that differ in the two lateral dimensions (see Fig. 3A), with a typical resolution of 6 × 3 × 2 mm (elevation, azimuth, range). The measured volume resolution of the T4 scanner is 3 × 3 × 2 mm, at a range of 70 mm with a 2.5 MHz transducer, which is small enough to resolve such relatively thin structures as valve leaflets and endocardium.

On-line display and capture

The primary display for the T4 system is a 60 Hz 640 × 480 pixel monitor receiving the output of the 3D scan conversion system in real time. The display may be configured to include up to 16 simultaneous slices through the 3D data set, although the standard configuration is five slices as shown in Figure 4 (an apical scan). The two B-mode images on the right are physically perpendicular within the heart, as shown in Figure 5A. The top B-mode image in Figure 4 is a 4-chamber view showing a slice through two ventricles, two atria, and the valves between them. The bottom B-mode image is a 2-chamber view. These views are familiar to cardiologists, who presently achieve them sequentially by rotating the transducer. With matrix array ultrasound, they are simultaneous views from the same 3D scan. The two B-mode slices may be swept independently through the data pyramid.

Displayed on the left of Figure 4 are three C-mode slices, which are image planes parallel to the face of the transducer. These C-mode slices show cross sections of the left and right ventricles displaying their con-

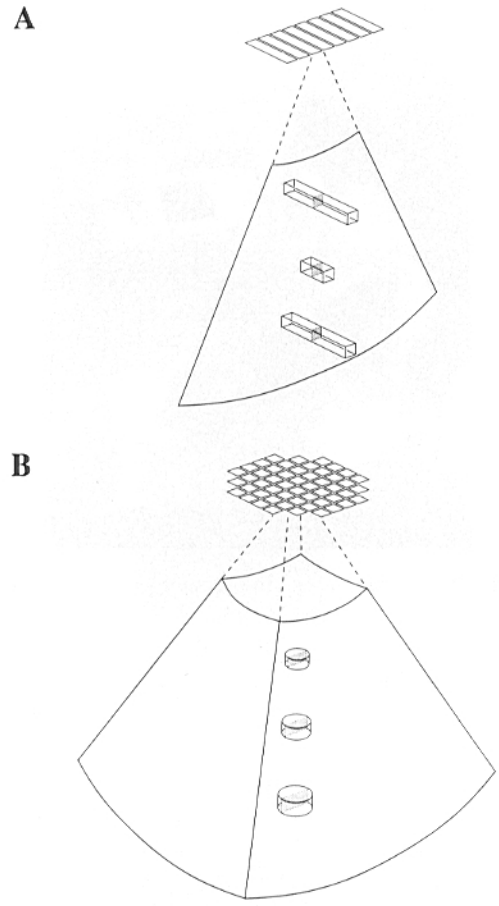

FIG. 3. (**A**) A linear (1D) array actively focuses in only one dimension, producing a voxel elongated in elevation through the plane of the slice. Although its cross-sectional area within the slice is small, its total volume is not. (**B**) A matrix array actively focuses in two dimensions, producing a symmetric voxel.

ventional short-axis shapes (i.e., round for the left ventricle [LV] and crescent-shaped for the right ventricle [RV]). To accentuate the 3D nature of the data, these C-mode slices are shown in projection (i.e., they are rotated into the screen). The location of the C-mode slices is indicated by arrowheads at the edge of the B-mode slices and may be changed by the operator. The left edge of the C-mode slices in Figure 4 corresponds to the left edge of the top B-mode slice, and the front edge of the C-mode slices corresponds to the right edge of the bottom B-mode slice. The thickness of the C-mode slices is adjustable. It should be emphasized that these slices are displayed simultaneously in real time while the transducer is held against the patient.

Because the matrix-array transducer is symmetrical, the two perpendicular B-mode images are interchangeable by physically rotating the transducer by 90°. There is a "handedness" in the display that dictates the right–left orientation of the bottom image relative to the top. By convention, we display a 4-chamber view on top, causing the 2-chamber view to appear reversed in terms of traditional right–left orientation. The transducer handle is marked with color coded dots that correspond to the arrowheads on the screen appearing in groups of three along the edge of each B-mode slice (colors not shown in Fig. 4).

Alternative displays available on the T4 scanner include two movable C-mode slices that can be tilted into trapezoidal shapes at arbitrary angles through the pyramid (see Fig. 5B). These inclined planes, or I-mode slices, are displayed without projection (see Fig. 6) and permit the operator greater freedom to move

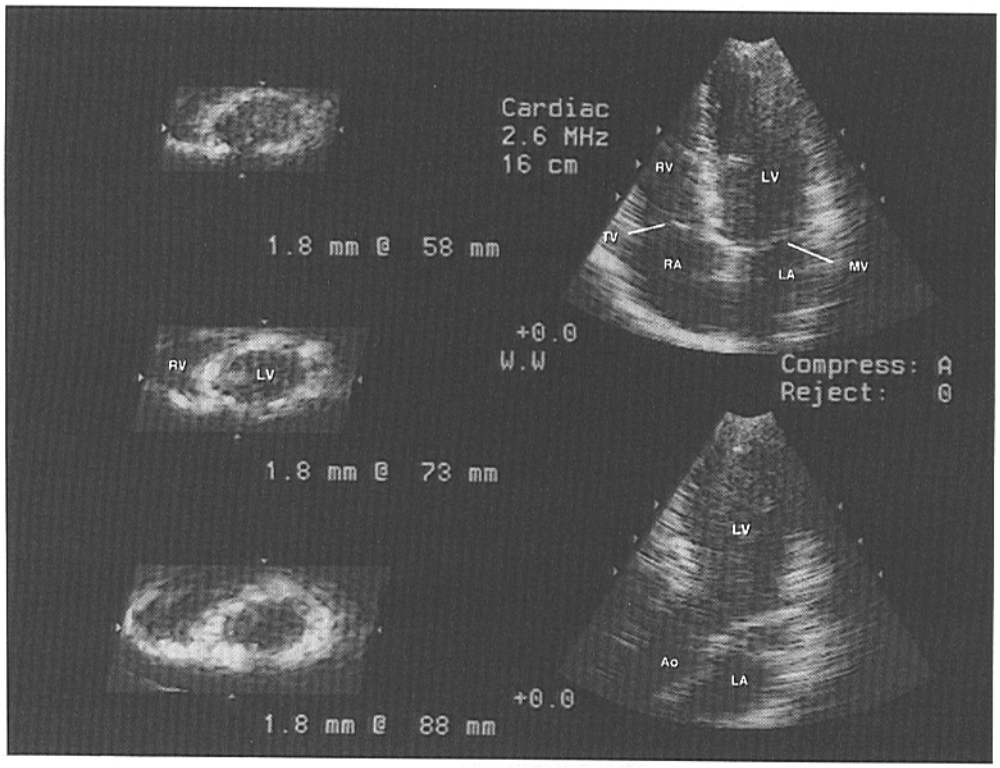

FIG. 4. Real-time display on the T4 scanner includes two orthogonal B-mode images and three stacked C-mode images in perspective. All slices may be moved in real time, and 3 seconds of stored data may immediately be replayed. LV, left ventricle; RV, right ventricle; LA, left atrium; RA, right atrium; Ao, aorta; MV mitral valve; TV, tricuspid valve.

into an optimal plane for diagnostic purposes. The white lines on the B-mode slices in Figure 6 indicate the location and angle of the two I-mode slices.

T4 produces a $64 \times 64 \times 512$ pyramid of 8-bit samples every 1/22 second, or 44 megabytes per second. The T4 scanner can record three seconds of data at this rate, enough to include at least one complete cardiac cycle. Recording is retroactive (i.e., the operator can record the previous 3 seconds at any time to capture singular events immediately after they occur). The complete 3D data set is stored, enabling all display functions to be reproduced. The stored data can be reviewed immediately on the scanner in a loop with interactive control of slice position, slice thickness, and playback speed, which in effect permits retroactive positioning of a virtual 2D transducer in 3D space. Alternatively, the data can be stored on an internal hard disk or transferred to a removable magneto-optical disk for archiving, off-line display, and analysis.

Off-line display

The quantity of information produced by the T4 scanner raises fundamental issues of optimal display for diagnostic purposes. We have developed software to provide off-line visualization capabilities on a high-performance graphics station equipped with hardware architecture known as *3D texture mapping*.[6] This specialized architecture permits rapid mapping from the 3D data set onto slices projected into visual space, providing capabilities beyond those of the real time display. The graphics hardware operates at a speed that allows interactive rotation, scaling, and slicing of the moving heart during replay, as well as interactive adjustment of the transfer function between voxel value and pixel brightness (compression curve).

Two methods of 3D rendering have been developed and tested for the replay of data from T4.

Intersecting slices. Two simultaneous orthogonal B-mode slices and one C-mode slice are combined into

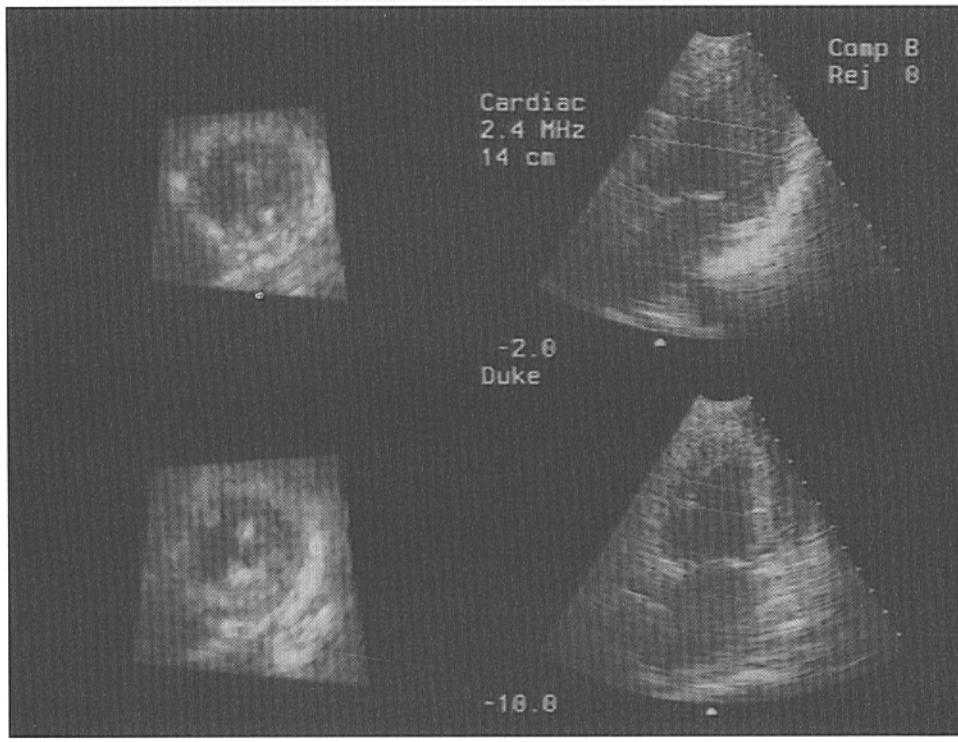

FIG. 6. Alternative real-time display on the T4 scanner showing two inclined C-mode, or I-mode, slices.

site direction would also be a long axis view). At the clipping plane, the "cut" tissue appears brighter compared to anatomical surfaces where the voxels are darker due to volume averaging. Figure 8B shows the data from the same scan rotated to a different view; this time we see a short axis view from within the LV. Here, the viewer is looking up from the left ventricle at the aortic valve and the mitral valve.

Figure 9 shows 20 successive frames at 22 frames/second of an apical scan, intersecting-slices, four chamber view from the posterior. The entire sequence represents just under 1 second of continuous data, approximately one complete cardiac cycle, as summarized in Table 1. It should be emphasized that this is not an average or gated study, but rather a single cardiac cycle captured in three dimensions in real time.

DISCUSSION

Matrix array compared to linear array

Conventional clinical ultrasound is constrained to a single 2D slice. This leads to ambiguities between in-plane and through-plane motion and requires the clinician to mentally reconstruct the 3D geometry of the heart. Previous methods of reconstructing 3D data sets by mechanically sweeping or turning a linear array are too slow to capture a single cardiac cycle in real time and require acquisition over multiple cardiac cycles while gating to the electrocardiogram.[7–20] Such methods suffer from transducer motion and respiratory motion, as well as nonperiodic cardiac motion. In the extreme arrhythmias make gated studies impractical. Even in normal patients, cardiac motion is not identical from one beat to the next. In the clinical setting, rapid variations in heart motion may be produced intentionally by drugs or other therapies, or may occur spontaneously. Monitoring beat-to-beat 3D cardiac function in the clinical setting has been impossible before the advent of matrix array ultrasound.

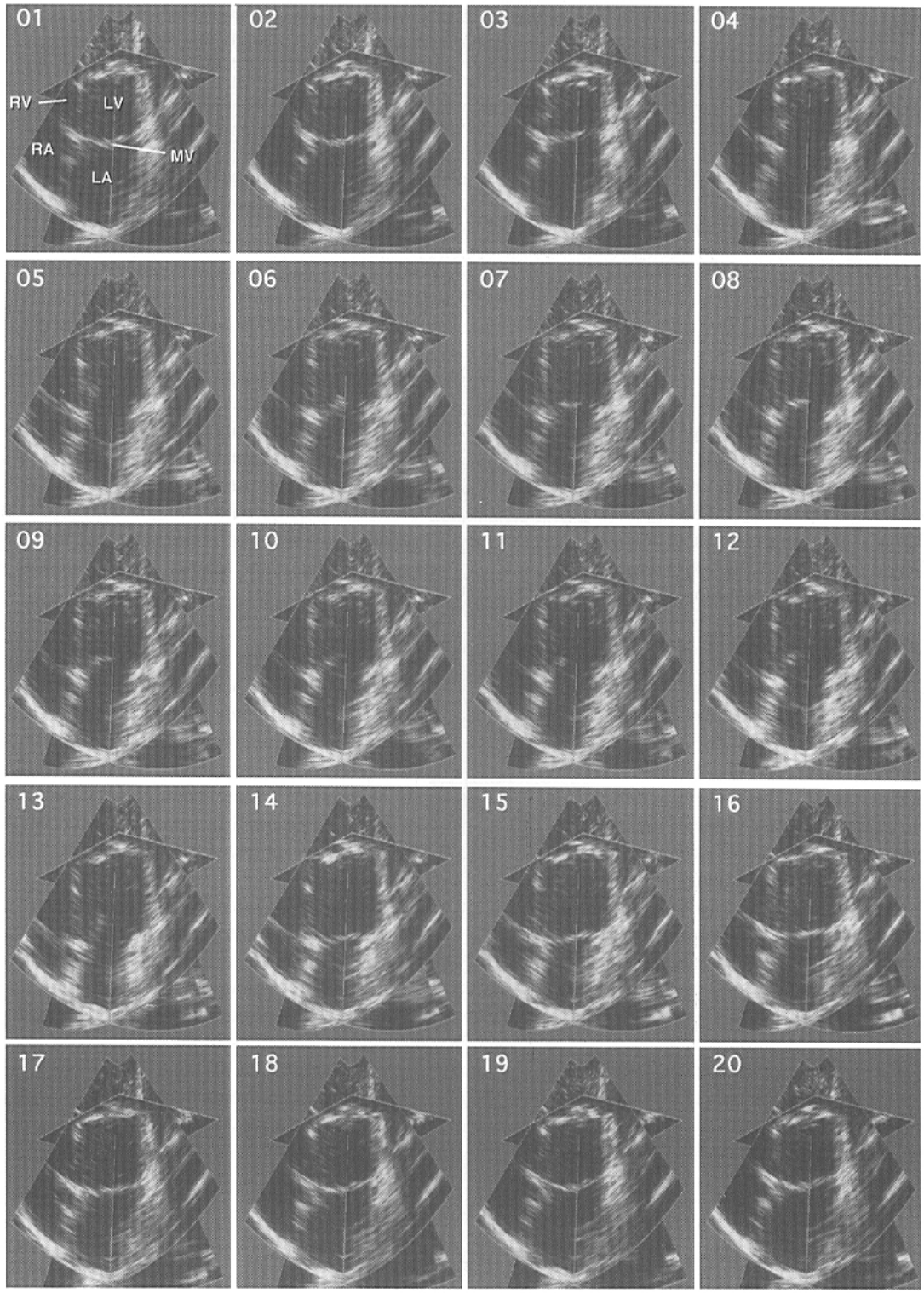

FIG. 9. Twenty successive frames at 22 frames/second of an *in vivo* human heart from a normal volunteer. A single cardiac cycle is shown extending from one systole into the next (see Table 1).

TABLE 1. EVENTS SHOWN DURING SINGLE CARDIAC CYCLE CAPTURED IN 3D AT 22 FRAMES/SECOND (SEE FIGURE 9)

Frame number	*Event*
1–3	End of systole
4	Mitral valve opening
4–9	Passive-filling portion of diastole
10–13	Active-filling portion of diastole
14	Mitral valve closing
14–20	Beginning of next systole

ary, as well as attenuation in the intervening path. Furthermore, the large amount of data presents challenges in its own right, and the frustrating fact remains that only a small fraction of the data can be displayed at once with meaning. Each B-mode slice contains only 1/64 of the entire data set, and each C-mode slice typically contains less than 1/50. As displayed on the T4 scanner, these slices must be viewed individually. Combining the slices into a geometric unit improves spatial comprehension, but slices are foreshortened and partially hidden behind each other. Even volume rendering, which begins with all the data, removes everything in front of the clipping plane and obscures much of what remains behind the opacity of bright surface voxels. Unfortunately, this opacity is necessary because the human visual system is simply ill-equipped to comprehend 3D translucent clouds.

Clinical significance

One of the most important practical advantages to 3D ultrasound is flexibility during replay. With 2D scanners, transducer position imposes a decision upon the data that the clinician cannot alter when reviewing it later. A playback station with 3D volumetric data permits the clinician to choose the location and orientation of slices during playback, reducing the job of scanning to simply positioning the heart within the scanned volume. This has important economic implications in terms of patient throughput. It reduces the time required to scan a patient and may facilitate the use of RT3D ultrasound in screening protocols (e.g., for congenital anomalies in children). It also has important implications for telemedicine, allowing a remote physician to explore anatomy without physically moving the transducer.

Although Doppler has not been included at the time of this writing, it is under development and will soon be available for RT3D ultrasound. The effect of Doppler on scan rate has not yet been determined, but is expected to entail the same compromise as seen in conventional ultrasound.

Matrix array ultrasound provides new opportunities to scientists and clinicians for analysis and diagnosis. Continuous direct measurement of cardiac chamber volume without geometric assumptions is now possible. Heart wall motion may be seen and analyzed directly, with implications in the diagnosis of myocardial infarction, cardiomyopathy, arrhythmias, and congestive heart disease. The ability to follow the injection of contrast agents, the results of electrical stimulation or other nonperiodic events may also prove valuable. The fact that 3D images are being captured synchronously means that a new kind of information can be derived, that of spontaneous shape.

CONCLUSIONS

Matrix array ultrasound permits an individual cardiac cycle to be captured in 3D with sufficient temporal resolution to study real-time cardiac motion. Many important structures in the heart are readily visible at the spatial resolution of our present data, and we expect the analysis of their shape and motion using RT3D ultrasound to have important clinical and scientific significance.

ACKNOWLEDGMENTS

Supported through a Whitaker Foundation Biomedical Engineering grant, NSF grant CDR8622201 NIH grants 1K08HL03220 and HL46242, with help from industrial partners in the NSF Engineering Center for Emerging Cardiovascular Technologies, Volumetrics Medical Imaging, Inc., Durham, NC, and Silicon Graphics, Inc.

REFERENCES

1. Shattuck DP, Weinshenker MD, Smith SW, Ramm OT. Explososcan: a parallel processing technique for high speed ultrasound imaging with linear phased arrays. J Acous Soc Am 1984;75:1273–1282.
2. Smith SW, Henry G, Pavy J, Von Ramm OT. High-speed ultrasound volumetric imaging system. Part I: Transducer design and beam steering. IEEE Trans Ultrason Ferroelectr Freq Contr 1991;38:100–108.
3. Von Ramm OT, Smith SW, Pavy HG. High-speed ultrasound volumetric imaging system. Part II: Parallel processing and image display. IEEE Trans Ultrason Ferroelectr Freq Contr 1991;38:109–115.
4. Von Ramm OT, Pavy HA, Smith SW, Kisslo J. Real-time, three-dimensional echocardiography: the first human images. Circulation 1991;84(Suppl II)685.
5. Smith SW, Pavy HG, Miglin JA, Ramm OT. Improved real-time volumetric ultrasonic imaging. Ultrason Imaging 1992;14:186–211.
6. Haeberli P, Segal M. Texture mapping as a fundamental drawing primitive. In: M.F. Cohen, C. Puech, F. Sillion, eds. Eurographics, Fourth Eurographics Workshop on Rendering, Paris, June 1993:259–266.
7. Kuroda T, Kinter TM, Seward JB, Yanagi H, Greenleaf JF. Accuracy of three-dimensional volume measurement using biplane transesophageal echocardiographic probe: in vitro experiment. J Am Soc Echocardiogr 1991;4:475–484.
8. Coppini G, Poli R, Valli G. Recovery of the 3D shape of the left ventricle from echocardiographic images. IEEE Trans Med Imag 1995;14:301–317.
9. Siu SC, Rivera JM, Guerrero JL, et al. Three-dimensional echocardiography: in vivo validation for left ventricular volume and function. Circulation 1993;88:1715–1723.
10. Aakhus S, Maehle J, Bjoernstad K. A new method for echocardiographic computerized three-dimensional reconstruction of left ventricular endocardial surface: in vitro accuracy and clinical repeatability of volumes. J Am Soc Echocardiogr 1994;7:571–581.
11. Pearlman AS. Measurement of left ventricular volume by three-dimensional echocardiography—present promise and potential problems [editorial; comment]. J Am Coll Cardiol 1993;6:1538–1540.
12. Sapin PM, Schroeder KD, Smith MD, Demaria AN, King DL. Three-dimensional echocardiographic measurement of left ventricular volume in vitro: comparison with two-dimensional echocardiography and cineventriculography. J Am Coll Cardiol 1993;22:1530–1537.
13. Jiang L, Siu SC, Handschumacher MD, et al. Three-dimensional echocardiography: In vivo validation for right ventricular volume and function. Circulation 1994;89:2342–2350.
14. King DL, Gopal AS, Keller AM, Sapin PM, Schroder KM. Three-dimensional echocardiography: Advances for measurement of ventricular volume and mass. Hypertension 1994;23:I(172–179).
15. Gopal AS, Keller AM, Rigling R, King DJ, King DL. Left ventricular volume and endocardial surface area by three-dimensional echocardiography: comparison with two-dimensional echocardiography and nuclear magnetic resonance imaging in normal subjects. J Am Coll Cardiol 1993;22:258–270.
16. Fisher JP, Wolfberg CA, Mikan JS, et al. Intracardiac ultrasound determination of left ventricular volumes: in vitro and in vivo validation. J Am Coll Cardiol 1994;24:247–253.
17. Hozumi T, Shakudo M, Shah PM. Quantitation of left ventricular volumes and ejection fraction by biplane transesophageal echocardiography. Am J Cardiol 1993;72:356–359.
18. Pandian NAG, Nanda NC, Schwartz SI, Fan P, Cao Q. Three-dimensional and four-dimensional transesophageal

echocardiographic imaging of the heart and aorta in humans using a computed tomographic imaging probe. Echocardiography 1992;9:677–687.

19. Greenleaf JF, Belohlavek M, Gerber TC, Foley DA, Seward JB. Multidimensional visualization in echocardiography: an introduction. Mayo Clin Proc 1993;68:213–220.
20. Belohlavek M, Foley DA, Gerber TC, et al. Three- and four-dimensional cardiovascular ultrasound imaging: a new era for echocardiography. Mayo Clin Proc 1993;68:221–240.

Address reprint requests to:
George D. Stetten, M.D., M.S.
Room 136, Engineering Building
Department of Biomedical Engineering
Duke University
Durham, NC 27706

Received September 1997; accepted with revisions November 1997