Linear Array Transducers with Improved Image Quality for Vascular Ultrasonic Imaging

This project not only achieved its goal of improving the near-field image quality of an existing transducer design, but also added two-frequency operation.

by Matthew G. Mooney and Martha Grewe Wilson

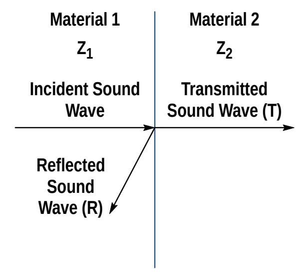
Medical ultrasonic imaging is a real-time technique that uses high-frequency sound waves to image many different parts of the body including the heart, vessels, liver, kidney, developing fetuses, and other soft tissue. The focus of this article is on noninvasive imaging of the blood vessels, which is more commonly referred to as vascular imaging. We begin with a general overview of ultrasonic imaging and then focus on the basic design aspects of a transducer used for imaging. Next, we examine the vascular market and the customer requirements, and then describe the design process used to develop two new vascular transducers. Finally, we present the clinical results.

The ultrasound waves used for imaging are generated by a transducer, which is held against the patient's body. The sound waves are produced by a piezoelectric material in the transducer. When a voltage is applied across the piezoelectric material, it deforms mechanically, creating vibrations in the material. These vibrations are acoustic (sound) waves and can have frequencies of 0.5 to 30 MHz. As the sound wave travels through the body and through various tissues, it bounces off the tissue interfaces, creating many reflections. The reflected sound waves are then detected by the transducer, providing information on the location of the tissues. A characteristic of the tissue called acoustic impedance determines the fraction of the energy that is transmitted from one tissue to another. The acoustic impedance, Z, is defined as:

$$Z = \rho c,$$

where ρ is the density of the material and c is the velocity of sound through the material.

Fig. 1 shows the sound wave at an interface and the transmission equation at that interface. In the body, the impedances are often very similar, so the system must be sensitive to these differences to distinguish blood from muscle, for example. As shown in Table I, the impedance of bone is much different, thus causing a great deal of reflection at that interface.[1,2] Since the sound waves cannot penetrate bone, natural "windows" in the body such as intercostal spaces between the ribs are often used. Attenuation, which is the result of absorption and scattering of energy within a material, creates another challenge in ultrasonic imaging.



$Z = \rho c$
where Z = Acoustic Impedance
ρ = Density of the Material
c = Velocity of Sound through the Material

Transmission Equation:

$$T = \frac{2Z_2}{Z_2 + Z_1}$$

Reflection Equation:

$$R = \frac{Z_2 - Z_1}{Z_2 + Z_1}$$

Fig. 1. Sound waves traveling between two materials.

Once the wave is reflected from the different interfaces, it travels back through the body and is sensed by the transducer piezoelectric material. The energy is then transformed into electrical signals which are propagated over a cable to the ultrasound imaging system, where all of the signal processing and image display occur.

Modern ultrasound systems are very complex and can consist of analog and digital portions. These systems process electrical signals in terms of amplitude and time, creating a real-time picture of the part of the body being scanned. Fig. 2 shows the HP Sonos 1000 cardiovascular ultrasound imaging system.

One type of ultrasound imaging system is the phased-array system. These systems have multiple channels for transmitting, receiving, and processing sound wave signals (64 and 128 channels are the most common). A typical transducer for these systems is divided into many individual transmitter/

Table I Characteristic Acoustic Impedances of Several Biological and Nonbiological Materials	
Material	**Acoustic Impedance (10^6 Rayls)**
Air at S.T.P.	0.0004
Water at 20°C	1.48
Blood	1.67
Muscle	1.70
Fat	1.38
Soft Tissue	1.63
Kidney	1.62
Liver	1.64
Bone	3.8 to 7.4
Polyethylene, Low-Density	1.8
Vinyl (rigid)	3.0
Lucite	3.2
Valox, Black (glass-filled nylon)	3.8
Aluminum	17.0
Lead Zirconium Titanate	28 to 36
Stainless Steel	45.4

Fig. 2. HP Sonos 1000 ultrasound imaging system.

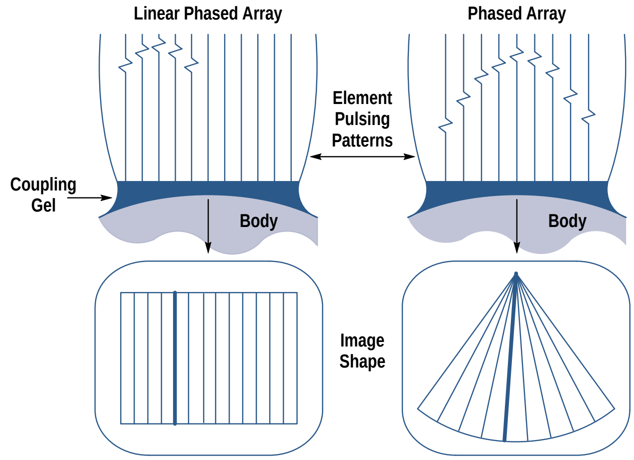


Fig. 3. Comparison of linear phased-array and phased-array ultrasound transducers.

receivers which are called elements. Typically, each transducer element is connected to one system channel. Two different types of transducers are commonly used: sector phased-array and linear phased-array transducers. The main difference is how these transducers are electrically excited. Each element in a sector phased-array transducer is activated at a slightly different time delay, which allows the sound wave to be shaped into a beam of sound and steered at different angles, producing a picture shaped like a pie slice. Linear phased arrays have the additional ability to activate groups of elements in a type of sequential scanning of the image, producing a rectangular-shaped picture. Fig. 3 shows a pictorial comparison of the element pulse patterns and the respective image shapes produced by the linear and phased-array transducers.

There are three main modalities in ultrasonic imaging: 2D, Doppler, and color flow. The 2D image is a real-time gray-scale image display. A typical 2D linear image of a carotid artery in the neck is shown in Fig. 4. Doppler is a way of measuring the flow velocity and movement within an image and is named for the principle it uses. The information is presented either with an audible tone or a visual plot. Color flow imaging detects the flow of blood and color-codes it depending on the direction and velocity of flow. An image showing color flow in the carotid artery is shown in Fig. 5.

The references provide more detailed information on phased-array ultrasound imaging systems[3] and on color flow and Doppler processing.[4]

Transducer Design

As mentioned above, a transducer consists of many small elements. An element is a multilayer sandwich of piezoelectric and other materials. The basic acoustic transducer element is shown in Fig. 6. Lead zirconium titanate, PZT, is a commonly used piezoelectric ceramic sensor material having an acoustic impedance between 28 and 36×10^6 kg/m^2s (Rayl). Recall that soft tissue has an impedance in the 1-to-2-MRayl region. Because of the impedance mismatch, there is a lot of reflected energy at the transducer/tissue interface, and such a transducer would not couple much energy to the human body. To improve coupling, a front matching layer—a coupling material with an intermediate acoustic impedance—is

Fig. 4. 2D linear phased-array image of a carotid artery.

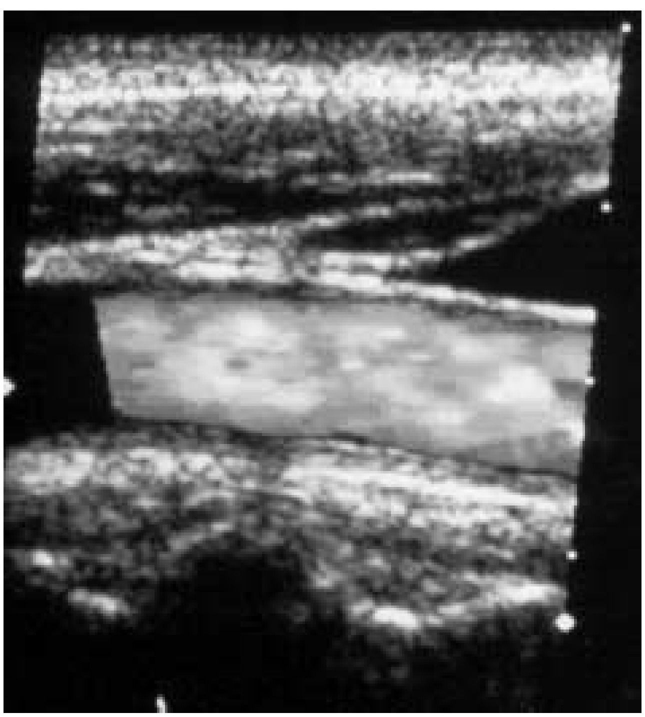


Fig. 5. Color flow image of a carotid artery.

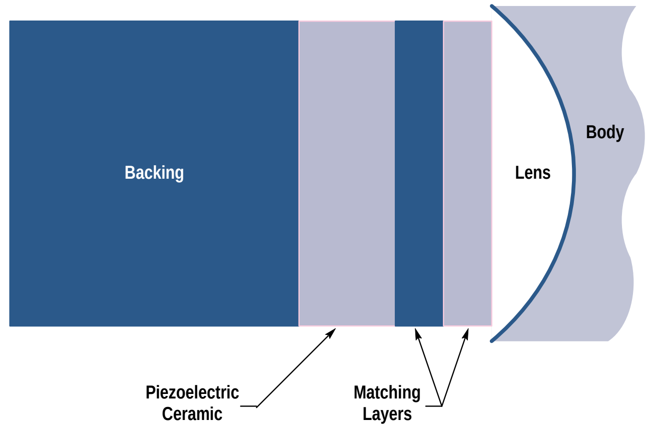


Fig. 6. Enlarged side view of one element of an ultrasound transducer.

added to the front face of the piezoelectric to aid in transferring the sound wave more efficiently to and from the body. An acoustically absorbing material called a backing is added to the back of the sensor to dampen energy that might cause additional mechanical vibrations (greater pulse length).[5,6]

An important aspect in ultrasound imaging is the ability to detect and resolve small structures in the body. This is largely determined by how well the beam of ultrasound is focused. Beam focusing for linear and phased arrays is determined by two different measures: elevation beam width and lateral beam width. Fig. 7 shows a picture of a focused beam and the elevation and lateral planes.[7]

The lateral beam width is a measure of transmitted beam width in the lateral plane and it changes as a function of many parameters including distance from the transducer. It can be measured by keeping the transmitted beam fixed while moving a receiving hydrophone in an arc in the lateral plane. Lateral beam width can be changed by electronically switching elements on or off to change the aperture size. The electrical impulses delivered to the elements can be advanced or delayed in time to provide additional focusing in the lateral plane. Fig. 8 shows a typical lateral beam plot at some distance from a transducer. Beam width is extracted from the beam plot and is a measure of how wide the main

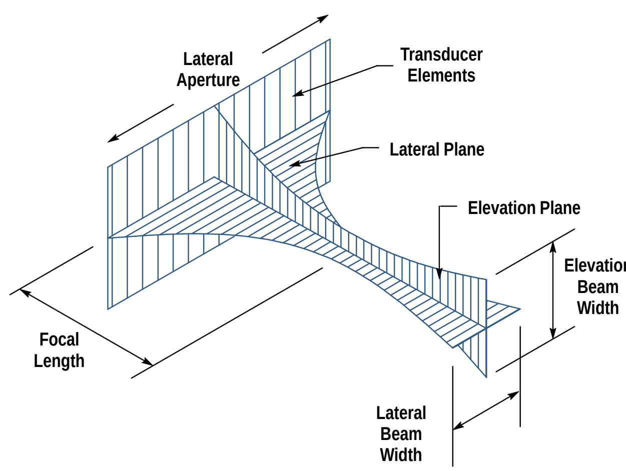


Fig. 7. Focused beam, showing elevation and lateral planes.

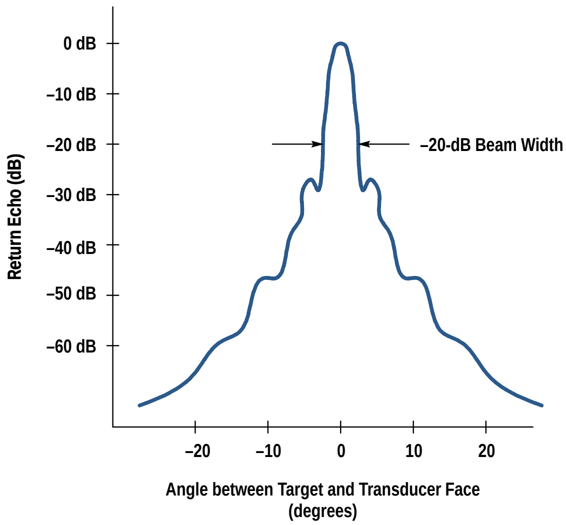
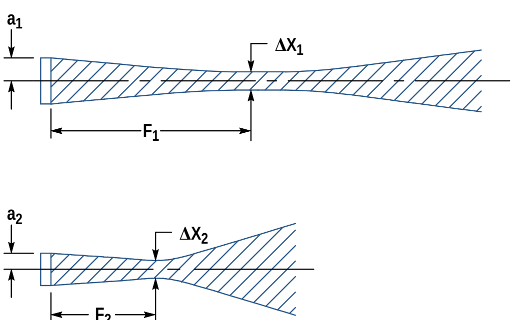


Fig. 8. Typical lateral beam plot at some distance from the transducer, showing beam width measurement.

transmitted lobe is at 6, 10, 20, or even 30 dB down from the maximum.

The elevation beam width is a measure of transmitted beam width in the elevation plane. Like the lateral beam width, it varies with the distance from the transmitting element. It is different from the lateral beam width because the size of the transducer in the elevation direction is not the same as the size in the lateral direction. Also, since most transducers are not divided into multiple elements in the elevation direction, elements cannot be electronically switched to change the size of the aperture, or phased to provide additional focus in the elevation plane. As such, the choice of elevation aperture is critical to any transducer design. Fig. 9 shows the effect of aperture size on elevation beam width and focal point.

Instead of electronic elevation focusing, a lens is placed over the elements to provide some focusing in the elevation



Formulas:[9]

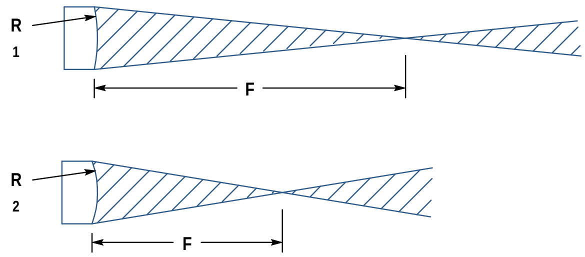
$$F = \frac{a^2}{\lambda} \quad (\lambda = 0.2 \text{ mm at 7.5 MHz})$$

$$\Delta X = \frac{0.89\ Z\ \lambda}{2a}$$

Where:
- F = Focal Length
- ΔX = Beam Width
- λ = Wavelength
- Z = Distance from Aperture
- a = Aperture / 2

	a	F	ΔX
1	10 mm	500 mm	4.5 mm
2	7 mm	250 mm	3.2 mm

Fig. 9. Effect of aperture size on elevation beam width and focal point.



Formula:[10]

$$F = \frac{R}{\frac{V_B}{V_L} - 1}$$

Where:
- F = Focal Length
- R = Radius of Lens
- V_B = Velocity of Sound in Body
- V_L = Velocity of Sound in Lens

	R	F
1	25 mm	50 mm
2	15 mm	30 mm

Fig. 10. Influence of lens radius on focal point and elevation beam width.

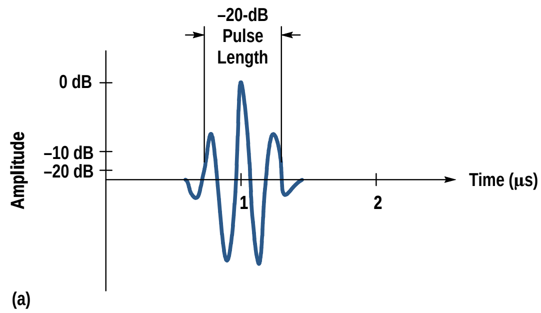
plane, but this focal point remains constant for a given choice of lens. The radius of curvature in conjunction with material properties of the lens determines where the narrowest point in the elevation beam will lie. This narrowest point is called the elevation focal point. The sketch in Fig. 10 illustrates how the lens radius influences both the focal point and the elevation beam width.

Another important factor in resolving small structures is time resolution. A well-focused beam allows small targets that are side by side to be resolved, while a short pulse length allows resolution of small targets that are separated by short distances into the body. Because the depth of an echo is determined by its time of arrival at the transducer, this attribute is called time resolution. Time resolution measures how well small structures can be resolved along the beam axis, so it is also called axial resolution. Axial resolution is measured in seconds of duration after excitation before the transmitted acoustic pulse fades to a certain level, usually 20 or 30 dB below maximum. The transducer can be designed to have a higher frequency, which increases axial resolution, but the body absorbs higher frequencies more, thereby reducing the depth into the body that can be imaged. Two pulses at two different frequencies are shown in Fig. 11. The higher-frequency transducer produces a shorter pulse. The choice of frequency requires a trade-off between axial resolution and depth of penetration.

Vascular Linear-Array Transducers

Since each imaging application has very specific requirements, different systems and transducers are sold for different applications. The medical ultrasound market can be segmented into cardiology, vascular, radiology, and obstetrics. Hewlett-Packard's product line is focused on the cardiology and vascular markets. Our project developed two transducers for the vascular ultrasound market.

In June 1990, HP introduced the HP 21258A 7.5-MHz linear phased-array transducer. This transducer offered several new technologies, one of which is continuous steering of the image. Since color flow and Doppler imaging are based on the Doppler principle, they are most sensitive when the blood



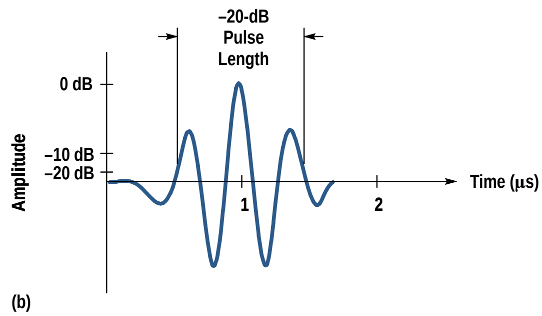


Fig. 11. Comparison of ultrasound pulses from transducers operating at different frequencies. (a) 7.5-MHz pulse. (b) 4.5-MHz pulse.

vessels and the blood they carry point towards or away from the transducer. Unfortunately, the 2D image has its best sensitivity when the blood vessels run parallel to the transducer. Steering can be done on the linear arrays to account for the conflicting angle requirements. This continuous steering is made possible by combining 288 transducer elements into 128 signal paths, which match the 128 channels on the ultrasound system. The ability to steer the 2D image to get the best grayscale picture while independently steering the color flow image to get the best color filling allows the user to get the best of both modalities. 2D image steering is an important feature of HP linear array transducers.

The introduction of the HP 21258A was followed in April 1991 by the lower-frequency HP 21255A 4.5-MHz linear-array transducer. The HP 21255A employed the same technology as the HP 21258A, but the lower-frequency ultrasound energy of the HP 21255A could penetrate deeper into the body. As expected, the axial resolution was not as good as the HP 21258A, since the frequency was lower. Together, these two linear array transducers provided the vascular ultrasound user with alternatives to the penetration/resolution trade-off.

As with any new product, improvements were planned for the next version soon after the introduction of the version A transducers. The improvements were centered around customer feedback and were organized into two major categories: the large size of the transducers and the inability of the HP 21258A to resolve small structures close to the surface of the skin (near field).

A cross-functional project team was formed to define and develop new versions of the version A transducers. The version B vascular transducer team soon divided the work to be done into two categories to match customer needs: ergonomic improvements and near-field image quality improvements. The remainder of this article will discuss the process used to address the near-field improvements.

Customer Feedback

The version B near-field image quality team set as an initial goal to be able to produce near-field images as good as the best competitor while maintaining our ability to produce good images in the far field.

Knowing that the near field needed improvement was a good start, but the project team required more detail. What did the customer define as near-field? Could this improvement be made at the expense of other attributes? How much improvement was needed? In addition to finding out exactly what the customer wanted, the project team also needed to figure out how to improve the near field. Should the frequency be changed? The aperture size? The focal point?

At this point in time, we needed a framework for organizing the information we had as well as some way to highlight areas that required more data. An organizational tool called quality function deployment (QFD) was identified as one way to help us translate what the customer wanted into a product. The QFD method is based on the construction of a "house of quality."[8]

The first step in building a house of quality is to tabulate customer wants and assign them weighing factors.[8] Given enough initial feedback, our team was able to develop and distribute a survey that listed many customer "wants" and asked for relative weights for each. The next step in building the house of quality is to tabulate those engineering characteristics that affect some or all of the customer wants. This list contains all the parts of the design that engineering could modify to give customers what they want.

The final steps in the housebuilding process took the most time. The competition's products were benchmarked relative to HP's to see in what areas of customer wants we were winning or losing. The engineering characteristics were related to customer wants in matrix form so that we could see what characteristics affected which customer want and by how much. The final house of quality provided a graphical means of displaying all this data in a readable format. A small piece of our house is shown in Fig. 12. The actual house had 20 wants and 30 characteristics and several "rooms" like the example in Fig. 12.

Engineering Design

The house of quality was an effective tool to show what we had to do to give customers what they wanted. Next the project team needed to change the design of the HP 21258A transducer to create the HP 21258B.

Using customer data, other rooms in the house of quality showed that the HP 21258A transducer was approximately equivalent to its competitors in the areas of color flow and Doppler performance. However, it was inferior to its competitors in near-field 2D image quality. The house of quality also showed that elevation beam width was the engineering characteristic most strongly related to 2D image quality. This became the first area of redesign on the HP 21258B.

The house of quality provided information that our customers considered the near field to be over a very specific depth range into the body. New HP 21258B designs were built and evaluated for elevation beam widths in this range. The HP 21258B designs included combinations of smaller elevation

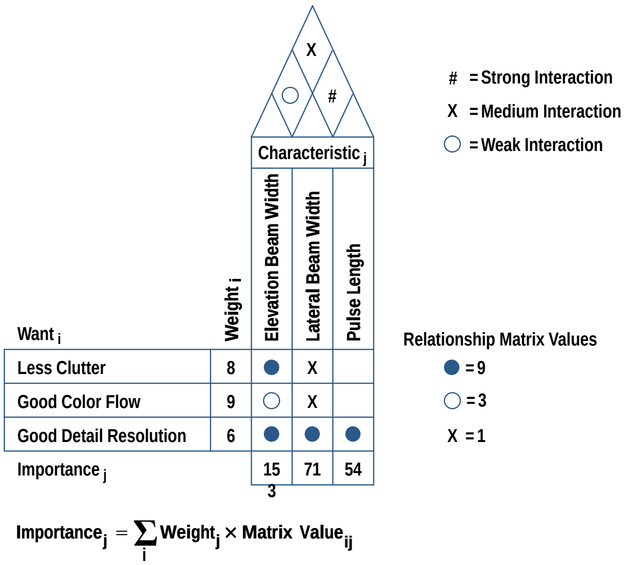


Fig. 12. A small portion of the QFD (quality function deployment) "house of quality" for the vascular ultrasound transducer.

apertures and tighter lens radii. The graph in Fig. 13 shows the elevation beam width of two such designs compared to the HP 21258A.

At least eight different designs for the 21258B were built and tested in terms of the critical few engineering characteristics shown to be important to customer satisfaction. A few of the most promising designs were taken to preference trials and tested against the competition.

Initial Clinical Trials

The initial preference clinical trials were held in-house with internal experts and some invited outside vascular ultrasound users. Images were acquired using the various transducer designs and customers were asked to grade the images relative to each other. Some preference trials were conducted at customer sites. The initial comments on the new designs were very positive from users of the HP 21258A. According to these users, the initial HP 21258B was definitely an improvement over the HP 21258A, especially in the near field. Unfortunately, the HP 21258B designs did not fare well in the eyes of those users that had experience with our competition. One technologist summed up the best of our new designs as "better than the old one … about 80% as good as my brand X."

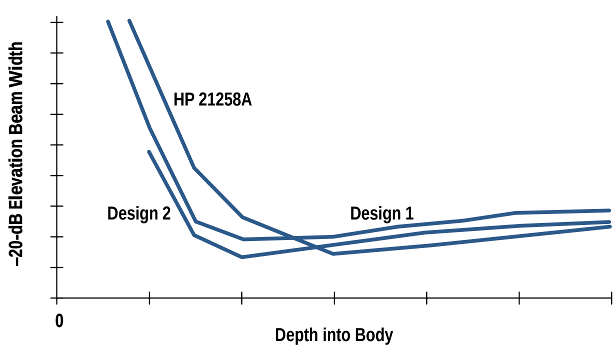


Fig. 13. Elevation beam widths of the HP 21258A transducer and two improved designs.

The disappointing performance of the initial HP 21258B designs required reexamination of the engineering design to identify further opportunities for improvement. The next most important engineering characteristics as defined by the house of quality were not transducer design changes but ultrasound system changes involving lateral beam width. There were some ongoing investigations into system improvements, but with the preference trial results, these efforts received more attention.

In terms of transducer improvements, many of the secondary engineering characteristics such as pulse length were shown to be moderately important to customer wants. Not one but many characteristics would have to be changed to make large improvements in transducer performance. Since we had already been through preference trials, we were pretty far along the product development cycle and a redesign would have taken too much time. At this point, some technique to change the transducer design significantly in a short period of time was needed.

Second Matching Layer

It was clear that an improvement in the axial resolution was required. A decrease in the pulse length would help. After investigating different ways to improve the pulse response of the transducer, one idea was to add a second matching layer to the sensor stack design.

Matching layers are very important to the transducer construction. The matching layer is attached to the front face of the sensor material and its main function is to help efficiently couple the energy to and from the body. A matching layer is one quarter of a wavelength thick to provide for constructive interference as the sound waves travel through it. By adding more than one matching layer, the energy is even more efficiently coupled, resulting in reduced pulse length. This would correspond to an improvement in the axial resolution.

The investigation of adding a second matching layer began by using computer models that showed a reduction in the pulse length when a second matching layer was added to a sensor. A comparison of the two modeled 7.5-MHz designs is shown in Fig. 14. The next steps in the process were to define the desirable material properties for the second matching layer material and then select and test an appropriate material.

In terms of transducer design, there are two important material properties of a matching layer: the acoustic impedance and the attenuation. Typically, the preferred acoustic impedance of the matching layer is the geometric mean of the impedances of the two materials it is sandwiched between. For example, if the first matching layer has an impedance of 8 MRayls and the body is 1.5 MRayls, the desirable impedance of the second matching layer would be 3.5 MRayls:

$$Z_{ML} = (Z_1 Z_2)^{0.5} = (8\ \text{MRayls} \times 1.5\ \text{MRayls})^{0.5}$$
$$= 3.5\ \text{MRayls}.$$

Ideally, the attenuation should be low to minimize the amount of energy lost when the wave travels through the matching layer.

In addition to the acoustic requirements, other material properties of the second matching layer were important. The material needed to be bondable since it was going to be

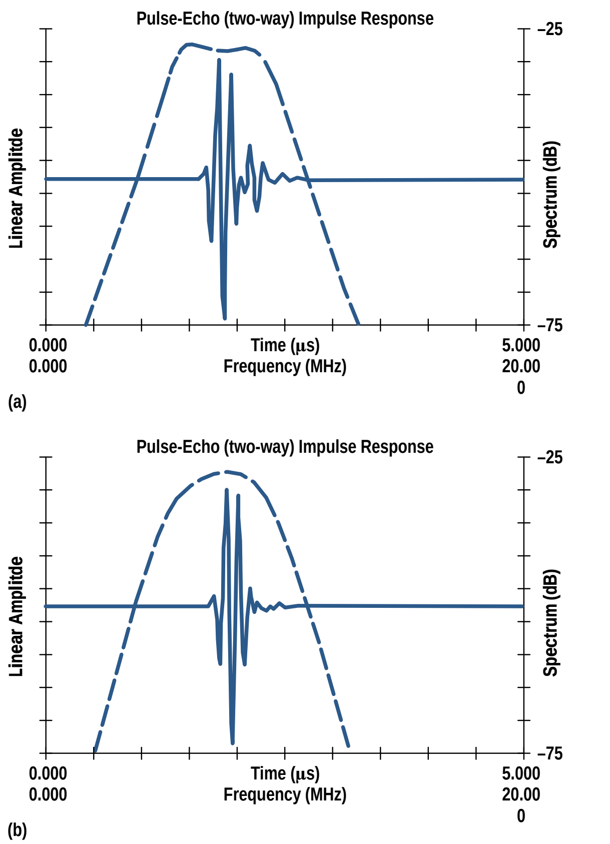


Fig. 14. (a) Pulse and spectrum of the modeled HP 21258A 7.5-MHz transducer. (b) Pulse and spectrum of the modeled HP 21258B 7.5-MHz transducer.

attached to the acoustic stack. It required good resistance to changes in temperature and humidity and resistance to chemicals to which the transducer would be exposed, such as acoustic coupling gel and disinfectants. The second matching layer also needed to be an electrical insulator for patient safety. To maintain consistent performance, each of the material properties needed to be homogeneous within the second matching layer.

Polymers generally have impedances between 1 and 4 MRayls as noted in Table I. To meet the acoustic requirements, many polymers were evaluated and the selection was narrowed based on the acoustic criteria. Next, more complete material property testing was done and a final polymer material was chosen.

To ensure that the second matching layer material selected was appropriate in terms of transducer reliability, extensive environmental testing was done. The testing included strife (stress to failure) testing in severe temperature and humidity environments. As a result of the strife testing, design changes were implemented to improve the robustness of the product.

A comparison of the acoustic response of transducers with and without second matching layers was done. The results showed that the pulse lengths are greatly reduced with second matching layers. Acoustic pulse and spectrum test data comparing 7.5-MHz transducers with and without second matching layers is shown in Fig. 15. The graph shows that there is a significant decrease in the –30-dB and –40-dB pulse lengths of the transducers with second matching layers.

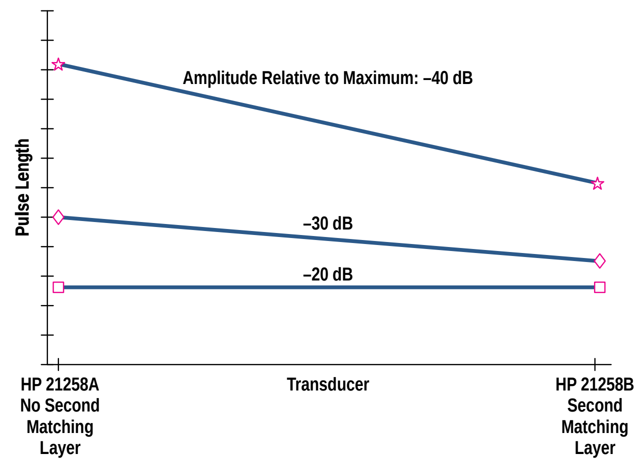


Fig. 15. Comparison of pulse lengths at various transmitted beam amplitudes below the maximum for HP 7.5-MHz transducers with and without second matching layers.

The main drawback to adding a second matching layer is acoustic cross-coupling between neighboring elements, which causes energy to be lost when the array is steered off-axis (roll-off). From the investigation, we found that the roll-off at 45° did increase slightly with the second matching layer. To gain a better understanding of how this would affect the 2D image quality and the color flow and Doppler performance of the transducer, clinical trials were done.

Further Clinical Trials

Several preference clinical trials were done in-house and at many different sites, both vascular and mixed cardiac/vascular. The main goal was to test the new version B linear arrays with added second matching layers against the version B linear arrays without second matching layers. In addition, clinical information was gathered on the performance of these new version B linear arrays versus the version A linear arrays and the competition's linear-array transducers. The two main areas of interest were the near-field resolution and the color flow performance.

The clinical trial results were very positive. A clear improvement in the 2D image quality was seen in all of the tests in which the version B transducer with the second matching layer was compared to the version B transducers without second matching layers. An example is shown in Fig. 16, which shows two 2D images of the radial artery in the arm (0.5 cm deep). The first image was taken with the version A 7.5-MHz linear array and the second was taken with the version B 7.5-MHz linear array. The near-field image quality is much improved with the version B transducer, as demonstrated by the clarity of the horizontal artery near the top of the righthand image in Fig. 16. The color flow and Doppler performances were determined to be about equivalent or slightly better. The version B transducer with the second matching layer also performed well against the competition's transducers.

Fig. 17 shows a bar graph comparing the clinical performance of the version A and B 7.5-MHz transducers with two competitors; the higher score indicates better performance. Overall, the near-field performance was improved, thus achieving the design goal.

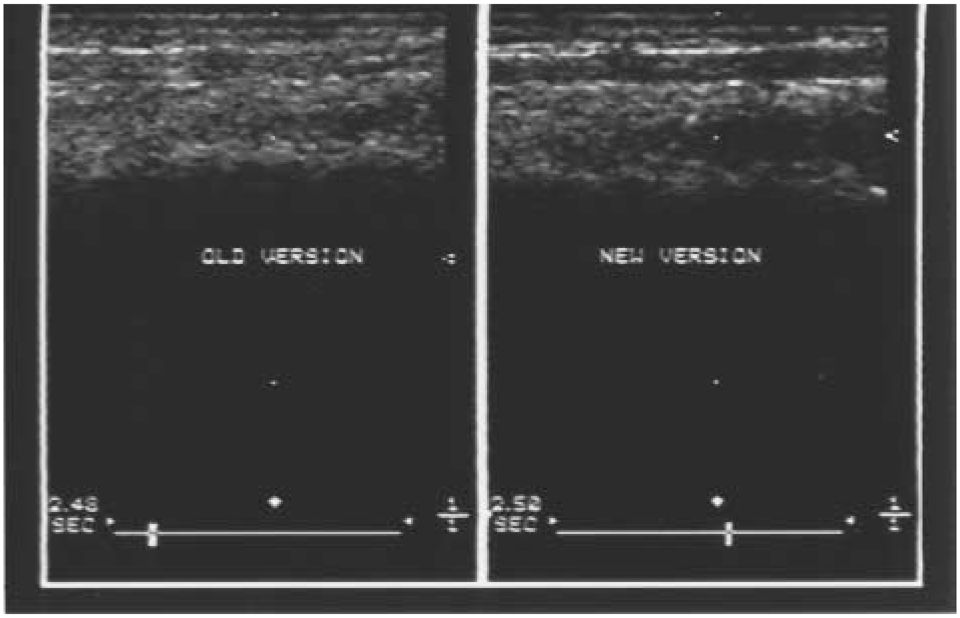


Fig. 16. 2D images of the radial artery in the arm. (left) HP 21258A transducer. (right) HP 21258B transducer.

Additional Features

In addition to the shorter pulse lengths, the new linear arrays have the ability to operate at two frequencies as a result of the increase in their bandwidth. With some system changes, the HP 21258B became a 7.5/5.5-MHz transducer and the HP 21255B became a 4.5/3.7-MHz transducer. This dual-frequency feature enhanced the performance of these transducers in addition to making them unique at that time in the vascular imaging market.

Manufacturing methods were also improved for the version B linear phased arrays. Taking advantage of state-of-the-art assembly techniques in one manufacturing step resulted in a threefold decrease in cycle time while making the process easier for the operators.

Another feature that was added was the ability of the transducers to image an expanded view. This new imaging format is called trapezoidal imaging and shows a much larger area. Trapezoidal imaging was developed outside the transducer image quality improvement project and is not covered in detail in this article. A comparison of a typical linear array image and a trapezoidal image is shown in Fig. 18. The expanded view allows sonographers to see more of the larger structures such as the kidney all in one image. Feedback on this new feature has been extremely positive. Customers have reported better and faster diagnoses with trapezoidal imaging. The trapezoidal imaging format is another distinct feature of these new linear-array transducers.

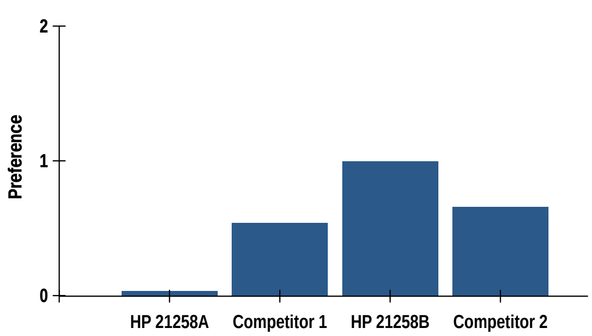


Fig. 17. Clinical trial results showing measured user preference for image quality of vascular performance of approximately 7-MHz transducers. Preference is measured relative to the HP 21258A.

The ergonomic portion of this project was also very successful. The size of the version B transducer was reduced by more than 25% compared to version A. New cable technology allowed smaller and lighter cables, resulting in a version B assembly that is two-thirds the weight of version A.

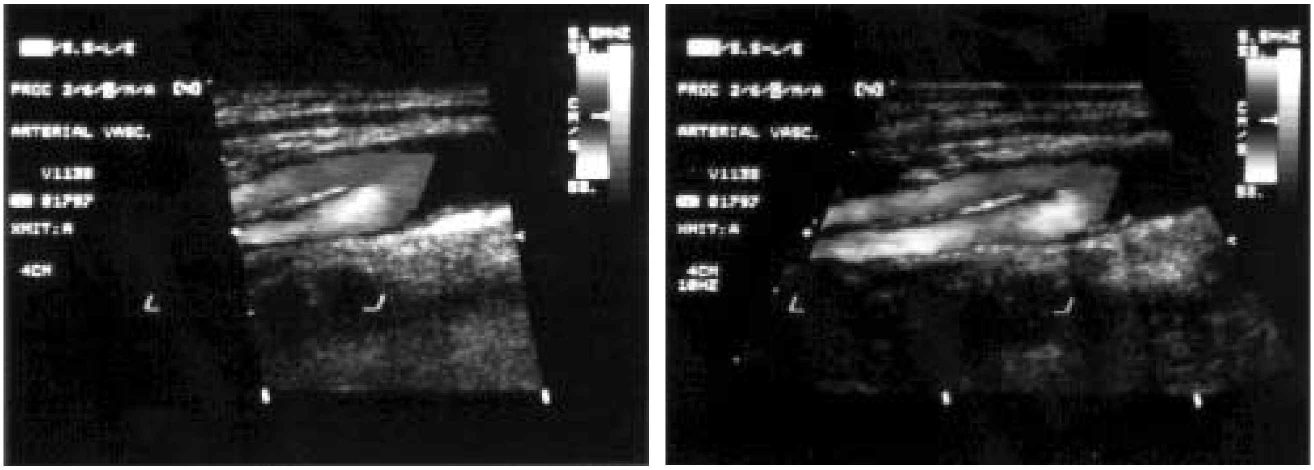
The version B vascular linear array transducers are shown in Fig. 19.

Conclusion

In June of 1993, the two version B linear phased-array transducers were introduced at the Society of Vascular Technology and the American Society of Echocardiology conferences. The transducers were well-accepted by physicians and sonographers, and the order rate for the version B linear phased-array transducers is several times that for the version A linear phased-array transducers. Today these new vascular transducers are helping clinicians provide more accurate diagnoses with improved image quality, improved ergonomics, and trapezoidal imaging format.

Acknowledgements

Many people contributed to the success of the transducer portion of the version B linear array project. Rick Snyder championed steerable linear arrays at the HP Imaging Systems Division in the early stages. Ray O'Connell was the R&D manager for this project. Some of the original image quality team members were Martha Keller, Gary Seavey, and Ed Parnagian. The ergonomics team members included Ed Parnagian, Reggie Tucker, Martha Moriondo, and Greg Peatfield. The teams had great technician help from Linh Pham, Jiuliana Ciarla, Sandy Decker, Tony Cugno, and many other manufacturing operators. The engineering teams were later expanded to include Greg Vogel, Hashi Chakravarty, Troy Nielsen, Sean Cranston, Paul Cartier, Jon Rourke, and Dan Stempel. A lot of help came from marketing which was represented by Pat Venters and Stockton Miller-Jones. Dan



(a) (b)

Fig. 18. 2D images showing the benefits of the trapezoidal format. (a) Linear-array format. (b) Trapezoidal format.

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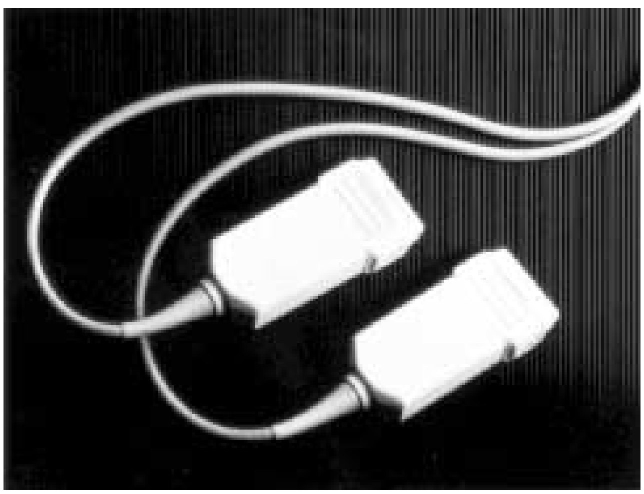

Fig. 19. HP 21255B and 21258B linear phased-array vascular ultrasound transducers.

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