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A Tissue-equivalent Upper Abdominal Phantom

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The first prototypes of an abdominal phantom have been constructed. The phantom is intended for eventual use in training diagnostic ultrasound personnel and in demonstrating commercial equipment. It is constructed from plastics believed to be stable and approximately tissue-equivalent at room temperature. Abdominal structures are formed from a dispersion of a polystyrene butadiene plastic in mineral oil. Polyvinyl chloride particles are incorporated to provide the desired attenuation coefficients and scattering levels. B-scans of the phantom produced realistic images, although problems associated with scanning technique and somewhat high phantom attenuation were noted. Very useful phantoms should result from relatively simple improvements in construction techniques. (Key words: tissue-equivalent phantoms; abdominal imaging; training phantom; tissue characterization; anthropomorphic phantoms)

In present ultrasound training programs, physicians and technologists learn the use of equipment by first spending many hours observing examinations and then practicing the imaging techniques on each other and on patients. This protocol is undesirable as it subjects the patient to possible discomfort and inconvenience as well as additional exposure to ultrasonic radiation. The process also is not an efficient use of the student's time.

This learning process would be improved and shortened by the availability of a tissue-equivalent phantom with anthropomorphic surface features and internal organs representing three-dimensional anatomy. With this phantom, the student could learn scanning techniques more efficiently, as he or she would be able to work continuously on improving a given technique using a reproducible portion of the anatomy represented in the phantom. With this additional experience, the student's first practice examinations of patients could be performed more rapidly and in a more professional manner. Furthermore, a realistic anthropomorphic phantom should be useful for demonstrations and comparisons of ultrasound systems and for sonographer examinations.

The earliest abdominal training phantom was developed by Holmes and Williams.[1] This system allowed the trainee some experience in imaging specular reflectors and simple scatterers but did not provide a realistic scanning surface or scattering regions as complex as those of the human body. The first attempts at simulation of scattering from parenchymal tissues involved polystyrene and glass microspheres in silicone rubber which did not have a tissue-equivalent attenuation or speed of sound propagation.[2,3]

Only recently have solid materials been developed which provide to a reasonable extent the attenuation, speed of propagation and scattering of ultrasound that is common in parenchymal tissues of human organs.[4–9] Most of these materials are water-based systems in which evaporation and organic growth must be controlled.[5–8] Use of these tissue-equivalent materials for instrument quality control, performance evaluation, imaging research, and design studies has been considered in many recent publications and presentations.[10–13]

Concurrently with the work reported here, a thin cylindrical training phantom was developed from essentially tissue-equivalent materials to sim-

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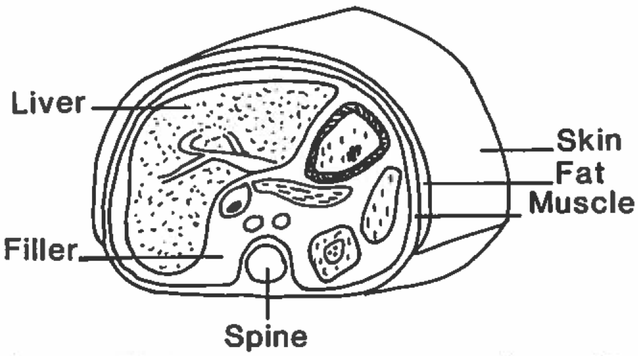


Figure 1. Sketch of the phantom construction showing levels of details of structures included.

ulate part of the human torso.[13] All but the scattering properties of the materials were shown to match those of human tissues. While not giving an anatomically correct image of the human abdomen, this phantom is useful in that it mimics several aspects of the human torso that produce artifacts in B-mode ultrasound scans.

The development of an abdominal phantom that has organs that are approximately anatomically correct in three dimensions and that also provides a realistic scanning surface is described below. This phantom was designed to give ultrasonic images as close as possible to those of the normal human abdomen with use of system control settings and scanning techniques employed routinely in clinical examinations.[14] To insure its utility for training, the phantom is designed to be ready for use with little preparation. It should require little or no maintenance and its properties should be stable over a period of five years, although variation of imaging properties with temperature is desirable to give added variety in training. A diagram of the desired configuration of the phantom is shown in figure 1. Not all of the various design goals have been achieved perfectly, but a remarkably functional phantom has resulted.[14]

MATERIALS AND METHODS

The copolymer used to form a majority of the internal structures of the phantom is Shell Oil Company's Kraton brand styrene–butadiene resin with mineral oil absorbed in the butadiene chains. The material was obtained as a liquid dispersion from the Research Division of 3M Company. The dispersion can be easily gelled to form organs by heating to 300° C, pouring into organ-shaped molds, and allowing to gel while cooling. When this technique is used, the copolymer will degas at the high temperature. Future availability of the 3M dispersion material is questionable, but essentially the same material can be obtained in a pregelled state from 3M's medical distributors as Flotation Pads® and melted for use with the high-temperature process described above.[4,9] In addition, a similar copolymer has been developed from the same material which has a lower, organ-equivalent, attenuation at frequencies to 3.5 MHz or greater.[15]

At room temperature, the gelled 3M material employed has an attenuation coefficient without scatterers of 0.22 dB cm^{-1} $MHz^{-1.75}$. That is, the attenuation increases with frequency as $f^{1.75}$. This value was obtained from measurements at frequencies of 1.5, 2.0, 2.5, and 3.0 MHz using displacement of water with a through transmission apparatus.[16] The speed of ultrasound propagation at room temperature (23° C) is approximately 1,455 m/sec as measured using a pulse transit time technique.[16] Density of the cast dispersion at 22° C is approximately 0.92 g/cm^2.

The gelled 3M copolymer is used without scatterers for anechoic regions such as the interorgan spaces, the subcutaneous fat layer, the interiors of blood vessels, and the gallbladder. The 3M material for the subcutaneous fat layer is molded to have the same tackiness as the 3M Flotation Pad, such that it will adhere easily to the urethane muscle layer.

Other organs are given increased attenuation as well as their scattering characteristics by incorporation of polyvinyl chloride (PVC) chips or powders of various particle sizes into the 3M material before molding. The density of PVC used is 1.32 g/cc. Of the 14 materials studied for possible use as scatterers, the PVC material had the best properties with regard to stability in accelerated aging tests, high scattering intensities achievable with relatively low added attenuation, and density reasonably close to that of the 3M polymer. Mean diameters of the particles employed in the approximately 25 scattering samples studied ranged from 60 to 280 μm. The density of PVC prohibited the use of particles larger than 500 μm in diameter, as these larger particles would not stay suspended long enough for gelling to occur. Recently, we have determined that cross-linked polystyrene particles have adequate chemical stability in the heated gel. With a density of 1.06 g/cm^3, much closer to that of the oleophilic copolymer, settling of the particles during organ cooling is not such a problem.[15]

Organs and vessels were separated or outlined both structurally and acoustically by coating each structure with an alcohol solution of polyvinyl butyral resin. In one of the three phantoms constructed, PVC scatterers were incorporated into this coating to provide a more echogenic outline. This coating also protected smaller components, such as vessels, from losing their shapes when hot 3M dispersion material was molded around them.

Urethane plastics from A.T.S. Laboratories were employed for the remaining components of the phantom. These plastics had ideal density and speed of ultrasound propagation and were more rigid than the gelled 3M material. They were more highly attenuating than most soft tissues, however,

Table 1. Composition of Scatterers in and Acoustic Properties of Abdominal Organs in the Phantom

Structure	Mean Diameter (μm)	Concentration (No./mm³)	Attenuation Coefficient (dB cm^{-1} MHz$^{-1.5}$)	Relative Scattering (dB*)
Liver	215	1.4	0.41	+2
Renal cortex	150	1.3	0.28	−9
Renal calices and stomach wall	280	0.32	0.48	+5
Spleen	100	1.1	0.26	−20
Gallstones	280	3.2	2.9	—
Pancreas	250	0.15	0.5	−5

* Scattering is given in dB relative to *in vivo* liver.

and were employed primarily for structural support components of the phantom. At 23° C, this plastic had a mean attenuation coefficient of 1.3 dB cm^{-1} MHz$^{-1.5}$ in the range of 1.5 to 3 MHz and a speed of ultrasound propagation of 1,540 m/sec. Thus, the 0.5 cm thick muscle layer added approximately 4.4 dB of attenuation to the phantom at 2.25 MHz.

Other A.T.S. Laboratory urethanes employed included highly attenuating urethane "chunks of food" in the stomach, a semirigid urethane foam, molded to simulate the diaphragm, and a clear, hard thermosetting urethane moded to form the end plates of the phantom.

The skin material used on the three phantoms was urethane sheeting, 0.005 inch thick, obtained from J. P. Stevens Corp. According to our measurements this material should add no noticeable attenuation to the phantom, while providing a smooth, flexible surface on which to scan. The spine was cast from a rigid polyester material.

Attenuation and Scattering Characteristics of Organs

The attenuation coefficient of the non-scattering components, the blood vessels, fat, interorgan material, and gallbladder, is that of the 3M gel, 0.22 dB cm^{-1} MHz$^{-1.75}$. At 2.25 MHz, this is 33 per cent less than the expected attenuation of 0.6 dB cm^{-1} MHz^{-1} for fat, while being high for the fluid-filled structures. As these fluid-filled structures were all small, the attenuation was not noticeable.

For the major organs, the attenuation was increased with the addition of scatterers to the 3M gel, giving the nearly tissue-equivalent or slightly high attenuation coefficients listed in table 1.[14] The composition of PVC scatterers in the copolymer, as chosen for the major organs, and the resulting backscattering at 3.5 MHz from these organ materials relative to that from liver in vivo are summarized in table 1. To better fit in vivo measurements, compositions of the spleen and pancreas should be changed to bring the backscattering to −10 and 0 dB, respectively, and to make a coarser texture in the pancreas.

The choice of scatterers was dictated mainly by the desire to achieve an accurate relative backscattering level (±6 dB) and texture for each of the organs. Scattering of human abdominal organs in vivo relative to that of liver was obtained by comparison of the pulse-echo sensitivity settings required to display liver scattering and the organ scattering with the same texture near the display threshold. A 3.5-MHz, 13-mm diameter, long-focus transducer was employed. The system sensitivity settings were calibrated with an rf attenuator. Since liver tissue exists at a wide range of distances from the transducers, scattering from other organs could be compared with that from liver tissue at the same range. Data were not taken where it was apparent that there were differences in the attenuations of overlying tissues.

Liver equivalence of the scattering from one of the samples was determined from liver scans of several lean patients and scans of the samples placed at the focal point (6 cm) in water. The same gray-scale scanning and 3.5-MHz transducer were employed for both sets of scans. Again, the relative system sensitivity settings were recorded, which allowed display of the scattering at a certain sparse texture near the display threshold for the scattering. Attenuation of the tissues overlying the focal point in the liver was estimated as 1.8 db cm^{-1} MHz^{-1} in the abdominal wall with 0.6 dB cm^{-1} MHz^{-1} in the liver overlying the focal point. The value 0.6 dB cm^{-1} MHz^{-1} is slightly larger than the most recently quoted value of 0.4–0.5 dB cm^{-1} MHz^{-1}.[17,18,¶]

Use of the lower attenuation coefficient of 0.45 dB cm^{-1} MHz^{-1} in scattering from the liver would indicate that the scattering from the "liver-equivalent" samples is approximately 6 dB too high.

Comparison of the scattering display threshold for the liver-equivalent sample with the display threshold for the echo from a perfect planar reflector indicated that the visual threshold for the liver-equivalent scattering at 3.5 MHz was approx-

¶ Ranelli R, Rohe Corporation, private communication.

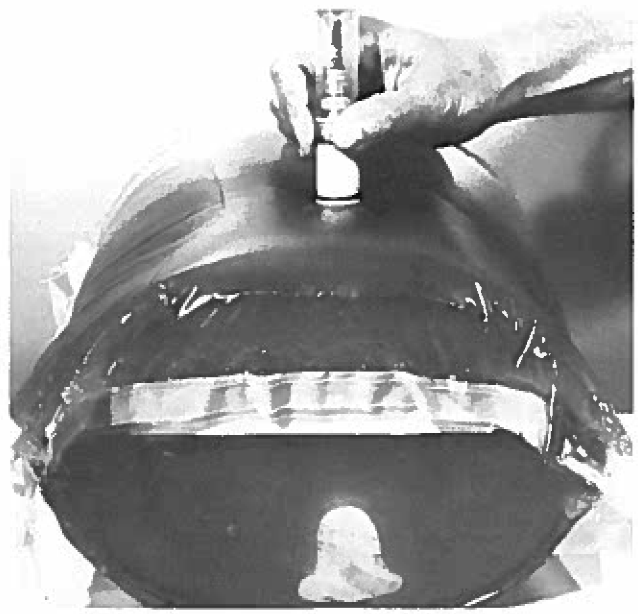


Figure 2. Caudad view of upper abdominal phantom being scanned. The replaceable skin and soft subcutaneous fat layer conform to transducer faces of various shapes even during sectoring.

imately 43 dB less than the *peak* amplitude of the echo from a perfect planar reflector.

More quantitative measurements of *mean* scattering amplitudes from the samples relative to that from a stainless steel plate were within +3 dB of the visual determinations of settings for threshold scattering.[19,20]

Structure of the Phantom

The phantom pictured in figure 2 was constructed as described above and illustrated in figure 1. Internal components of the phantom include:

1. *Liver*, formed of 3M olephilic copolymer with PVC scatterers and containing some hepatic and portal venous structures with no scatterers.
2. *Gallbladder*, formed with clear 3M copolymer and containing an approximately 0.52-cm gallstone simulated with the 3M copolymer containing a high concentration of scatterers.
3. *Stomach*, molded in two sections, an inner scatter-free area of copolymer containing urethane "chunks" to simulate "food" and an outer wall formed with copolymer containing PVC to give +5 dB backscattering intensity relative to the liver.
4. *Left and right kidneys* with simple collecting systems, all formed from the copolymer containing PVC scatterers to give a highly echogenic collecting system and weak echoes from the cortex.
5. *Pancreas*, molded of copolymer and PVC as described in table 1 with a backscattering intensity of −5 dB relative to the liver.
6. *Spleen*, composed of copolymer and PVC scatterers to give very weak echoes compared with the liver (−10 dB would be ideal).
7. *Major blood vessels* (aorta and inferior vena cava), formed from clear copolymer and running the full length of the phantom.
8. *Diaphragm*, a semirigid plate of urethane foam at the cephalad end of the phantom. The xiphoid process is defined at the caudad end of this foam plate as seen from the anterior end of the phantom.
9. *Spine*, cast of rigid plastic running the length of the phantom.

A layer of fat-equivalent plastic lies above the muscle layer and is held in place by a simulated skin layer, which is attached to the trunk on both sides with Velcro strips. The phantom can be set in a firm plastic case which is supplied to provide protection during storage and shipping. Mineral oil was employed as a filler in two of the three phantoms produced, in which the organs and interorgan dispersion were of inadequate volume to fill the urethane muscle shell.

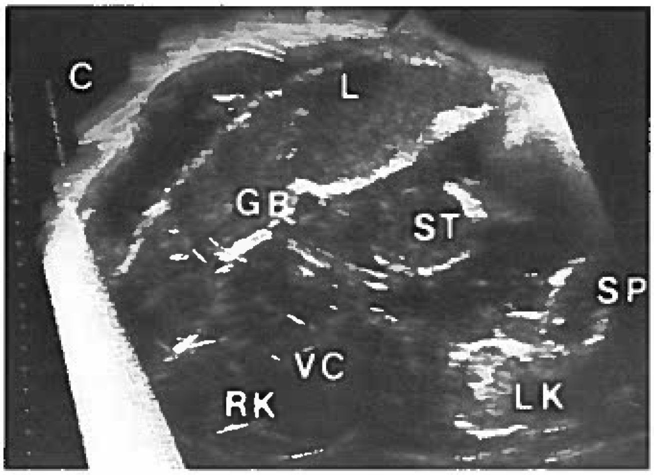
RESULTS

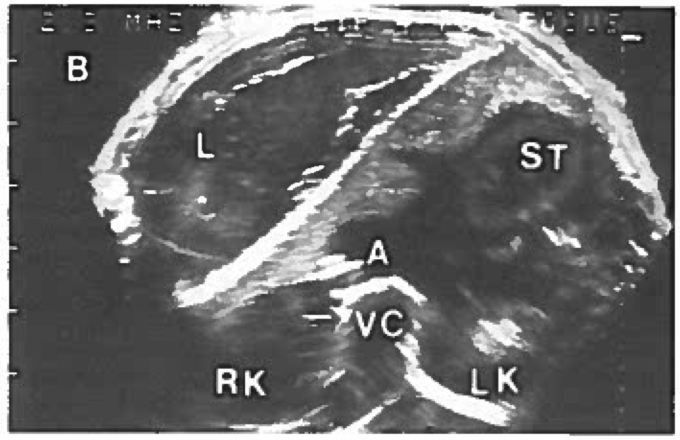
B-Mode Scans of the Phantom

B-mode scanning of the three models of the abdominal phantom was performed at the University of Colorado Health Sciences Center on digital and analog compound scanners. Images presented here were obtained using a 2.25-MHz, 13-mm-diameter transducer. The phantoms in their final form proved too attenuating to allow visualization of the posterior organs at 3.5 MHz, and skin friction made compound scanning more difficult with large-diameter transducers. Images were obtained by compound scanning over the anterior surface of the phantom, where the presence of the fat pad allowed the necessary flexibility. A full 180° arc could not be made on the model C phantom because of the presence of the glue holding the muscle layer of that phantom together at the sides.

Images discussed in this paper were obtained from two models of the abdominal phantom, Models B and C, which contained the same organs and blood vessels but differed slightly in their methods of construction. Similarities and differences in imaging qualities of the two phantoms are illustrative of the capabilities of such anatomic phantoms and provide a basis for decisions in the construction of more widely available phantoms. In Model B, identified by a B in the upper left corner of the image, major organs were coated with clear Butvar resin, while, for organs in Model C, scatterers were incorporated into the resin coating. In both phantoms, PVC scatterers were employed in coating blood vessels.

Figure 3 represents a transverse image of phantom C at a level 1.5 cm below the xiphoid process (as defined above). The liver (L), gallbladder (GB), stomach (ST), spleen (SP), left





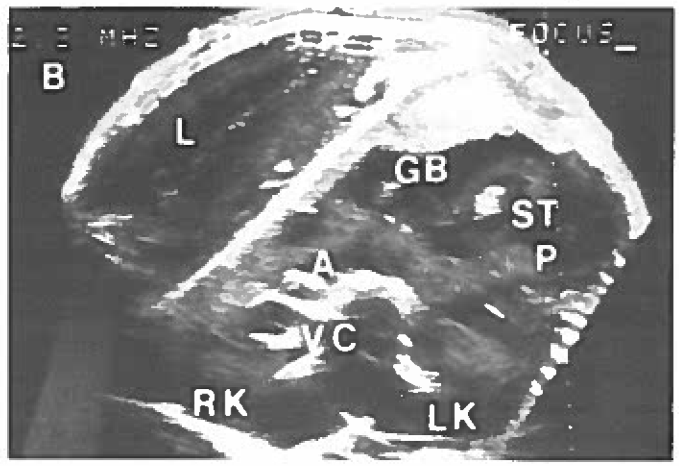


Figure 3 (*above*). Transverse image of phantom C with echogenic organ borders. The image level is 1.5 cm below the reference xiphoid process. See text.

Figure 4 (*right, top*). Transverse image of phantom B with non-echogenic organ borders. The image plane is 3 cm below the xiphoid process. See text.

Figure 5 (*right, bottom*). Transverse image of phantom B at a plane 8 cm below the xiphoid process. See text.

kidney (LK) and collecting system, cephalad tip of the right kidney (RK), and spine all are visualized at this level. The anterior and posterior walls of the vena cava (VC), also are demonstrated in this image. The echo-free region anterior and lateral to the liver is due to an excess of free mineral oil incorporated into the phantom. The highly echogenic borders of the organs are due to PVC scatterers incorporated into their Butvar coatings. Relative backscattering intensities compare well with the desired levels given in table 1, as the spleen appears nearly anechoic while the renal collecting system exhibits strong echoes when compared with the liver tissue backscattering.

The desired variation in backscattering textures of the organs as opposed to echo amplitudes is not evident in these scans performed at 2.25 MHz. Imaging at 3.5 MHz, as was performed on the sample organs, is necessary to begin to resolve any texture beyond the acoustic speckle common to all the organs containing small scatterers.

Figure 4 is an image of a transverse plane of a different phantom, model B, located approximately 1.5 cm caudad to that in figure 3 of Model C. Organs imaged are the same as in figure 3 with the exceptions that the gallbladder and spleen lie outside this image plane. No scatterers were incorporated in the Butvar organ coatings. Without the strong border echoes, organs of little or no backscattering intensity are not readily detectable unless they lie near other echogenic structures. In this phantom the anterior and posterior margins of both the aorta and the vena cava are imageable and the vertebral body stands out clearly. The intense line running through the liver parenchyma on all images of phantom B is due to the settling of scatterers in the two-mold process used at the time this phantom was constructed.

In the image of this phantom 5 cm further caudad (fig. 5) the pancreas is visible running from the phantom's left side under the stomach and over the vertebral body, ending near the liver. The backscattering amplitude of this organ approximates that of the liver, as desired, and there is just a hint of coarse texture. At this level, the collecting system of the right kidney is evident, as are the aorta and vena cava.

In phantom B the gallbladder is locatable only by the presence of gallstones. Certainly it would be desirable for the organ to have an echogenic border.

Presented in figure 6 is a series of sagittal scans in selected planes from left to right of the two abdominal phantoms with the cephalad end of the phantom to the left of the image. In addition to those organs illustrated in the previous scans, the

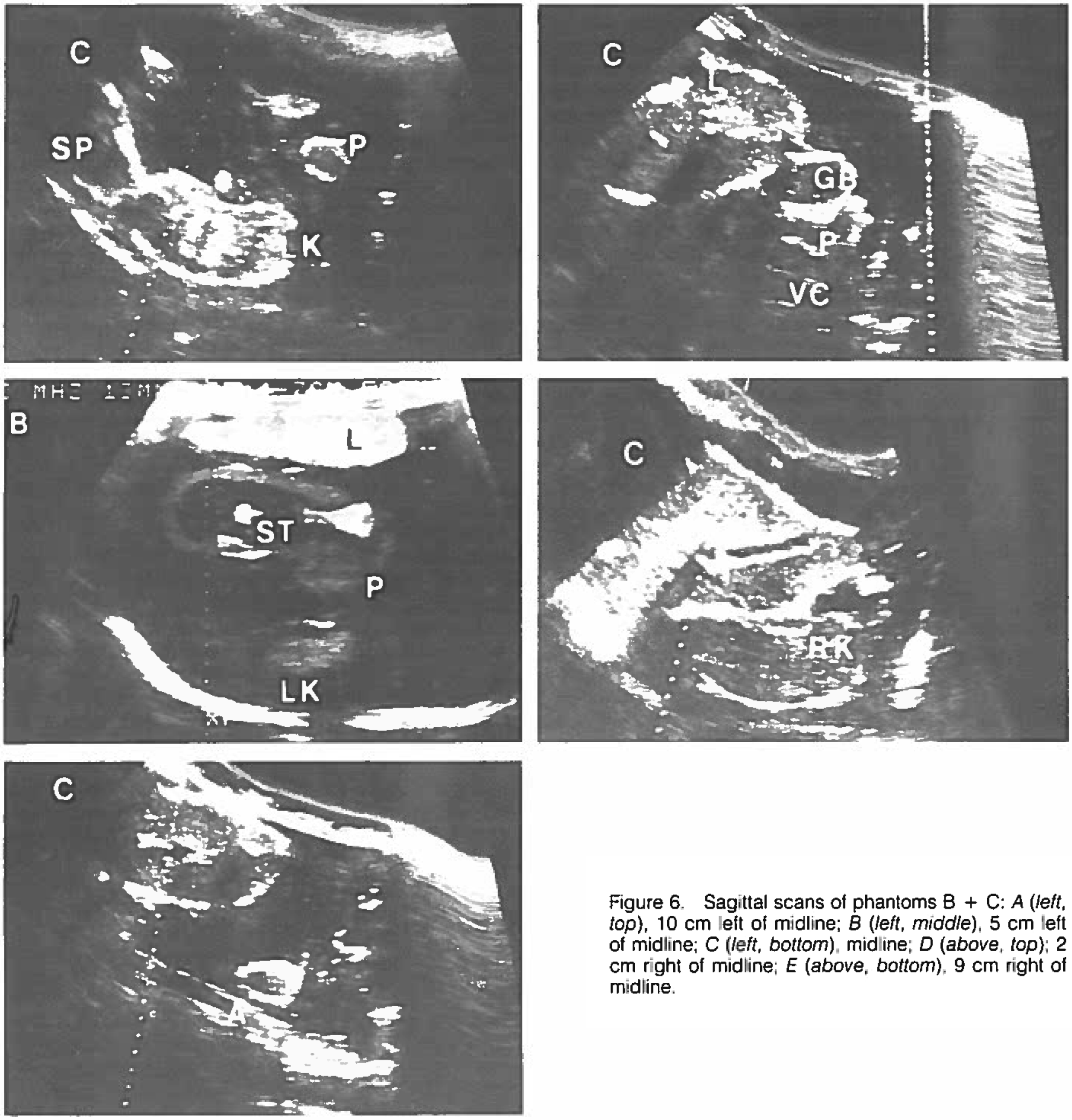


Figure 6. Sagittal scans of phantoms B + C: *A* (*left, top*), 10 cm left of midline; *B* (*left, middle*), 5 cm left of midline; *C* (*left, bottom*), midline; *D* (*above, top*); 2 cm right of midline; *E* (*above, bottom*), 9 cm right of midline.

hepatic and portal venous structures, ranging in diameter from 1 cm to 3 mm, are visualized. Shadows posterior to the portal vessels simulate effects encountered in vivo at higher ultrasound frequencies and result from attenuation by the high concentration of scatterers and possibly some air bubbles in the vessels.

DISCUSSION

The prototype phantoms described illustrate the feasibility of constructing human torso phantoms which are of value to clinical personnel for practice of scanning techniques and which are probably also useful for sonographer examinations and demonstrations of equipment. Scans of these models can produce images with realistic appearances and identifiable abdominal organs.

In general, all of the components of the phantoms were imageable and appeared with acceptable scattering intensities relative to one another. Since there were some anatomic imperfections and no scatterers were incorporated in the interorgan material, ultrasound images of the phantoms looked more like those of sick patients with ascitic fluid and slightly disoriented organs than like those of normal subjects. However, this did not detract substantially from the utility of the phantoms as training devices.

Novices as well as experienced sonographers were challenged by the process of adjusting controls, obtaining high-quality images of a particular anatomic location, and finding particular organs while scanning from one of the many possible directions of imaging. The genuinely three-dimensional shape and positioning of the various organs appeared particularly important in providing the necessary degree of difficulty and realism. Given the variety of techniques used in ultrasound scanning it is reasonable to expect that a phantom such as this will find much more prolonged use in ultrasound training programs than have anatomic phantoms in radiography and nuclear medicine training. To further challenge students, the phantom can be warmed or cooled, changing phantom attenuation and speed of sound characteristics for examinations.

Numerous gains in technical understanding were made to achieve the current phantom characteristics. Much of this new information concerns the backscattering characteristics of tissue. It is clear that low- to moderately-high-amplitude fine-texture scattering of organs can be reproduced without adding excessive additional attenuation by using PVC or similar plastic scatterers in the range of 100 to 280 μm in diameter. Associated with the scattering is an increase in attenuation over the attenuation of the base material which rises from negligibly small to unacceptably high at frequencies above 2.25 MHz. To obtain very-high-amplitude scattering such as that of the renal collecting system or to obtain an inhomogeneous pulse-echo appearance of scattering from certain tissues, it will be necessary to employ quasi-specular reflectors with dimensions on the order of 1 mm, or reflectors lying in curved sheets or in a regular spacing.

Limitations of the present phantoms include the comparatively high attenuation of the more strongly scattering organs, low speed of sound of the mineral oil copolymer material, and somewhat high friction and lack of long-term ruggedness of the skin and fat pad layers. The phantom's body wall also was molded in the shape of a standing person, so its A-P thickness is considerably greater and harder to penetrate than the A-P dimensions of a person of thin to medium build lying supine. The combination of phantom thickness and somewhat high attenuation necessitates the use of a 2.25-MHz transducer and a high system gain to obtain the penetration necessary to image posterior organs. This in turn results in a decrease in resolution and less than ideal texture in the images obtained, as well as not allowing the student experience in the use of scanner settings more consistent with typical patient examination.

Clearly, the thickness of the phantom can be reduced easily, and the availability of a formulation of the mineral oil–copolymer base material which has reduced ultrasound attenuation[15,21] will make more realistic imaging possible at 3.5 to 5 MHz as well as 2.25 MHz. If use of anechoic interorgan material is continued for ease of construction and reduced overall attenuation, the relatively hypoechoic organs such as the spleen should have echogenic borders.

The approximately 5 per cent deviation in speed of sound of the mineral oil–copolymer material from that of most average soft tissues may add to the degradation in resolution in the posterior regions of the phantom in compound scans. However, image degradation is not as bad as that obtained in scanning the large numbers of patients who have significant amounts of interorgan fat to refract the ultrasound beam.[13] The speed of sound of the phantom is quite homogeneous except for the thin muscle shell, which remains reasonably parallel to the transducer face. In the phantom, as in the body, travel time artifacts are small compared with refraction artifacts. For the phantom, travel time artifacts can be essentially eliminated by adjusting the time of the ultrasound instrument to correspond to the 1,455 m/sec speed of sound in the phantom. This adjustment is readily made on one American-made unit and is commonly available on most systems made outside the United States.

Characteristics of the outer layers on future phantoms should include the soft, replaceable subcutaneous fat layer and the slick, stretchy, replaceable skin. The subcutaneous fat layer needs to be made more rugged and secured to the phantom in a manner such that air cannot accumulate at the fat/skin and fat/muscle interfaces. The plastic skin should provide less friction for scanning with transducers of the larger diameters. Perhaps a vinyl or urethane skin with a Teflon-like coating would be ideal.

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