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REVIEW ARTICLE

Role of Intraoperative Doppler Ultrasound in Brain Surgery

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ABSTRACT

Early investigation and excision are required for brain surgical lesions. When a patient has brain lesion, surgery ought to be their primary option. Treatment and total resection of a brain lesion are nearly always achievable. Maximal safe excision of brain lesions requires accurate and dependable intraoperative neuronavigation. The next frontier in navigation improvement has drawn a lot of interest in intraoperative magnetic resonance imaging, or intraoperative Magnetic Resonance Imaging (iMRI). Unfortunately, most centers throughout the world are unable to use iMRI due to its prohibitive cost and practical difficulties. By contrast, intraoperative ultrasonography (ioUS) is a low-cost instrument that can be seamlessly integrated into the theater's current setup and operational procedures. In the past, ultrasonography has been thought to have poor, artifact-prone image quality and be challenging to learn and standardize. However, with significant advancements in image quality and well-integrated navigation features over the past ten years, ioUS has undergone a dramatic evolution.

Keywords: Doppler; Ultrasound ; Brain Surgery

INTRODUCTION

The fundamental principle of brain lesions is to maximize safe surgical resection in an effort to enhance overall survival, progression-free survival, quality of life, and symptoms. Accurately localizing tumors and distinguishing them from the surrounding functioning neural tissue without injury of blood vessels are still difficult tasks [**1**].

Planning surgical methods frequently involves the use of preoperative stereotactic imaging (MRI/CT). These systems, while strong instruments, are intrinsically restricted since they do not provide intraoperative real-time depictions of the tumor and surrounding structures. Their accuracy declines with further surgery because of erratic brain deformations, distortions, and shifts [**2**].

There is a chance of unintentional harm, which could result in a functional deficit, or leaving residual due to misjudged margins, which could affect the prognosis. These outcomes are caused by non-contemporaneous, imprecise navigation. As such, contemporaneous intraoperative imaging is clearly needed, as it provides an accurate map of the state of surgery today [**3**].

Ultrasound (US) is a repeatable, safe, and reasonably priced imaging modality that is simple to incorporate into surgical workflows, enabling live imaging during procedures. US has developed over the past 30 years as a neurosurgical tool and is now a standard procedure in many neurosurgical facilities **[4].**

In 1982, adult neurosurgery began using ultrasound (US) for the first time. At that time, 2-dimensional B-mode imaging (2D US) became available, allowing real-time viewing of neural architecture and pathology during surgical operations. Since then, surgical planning has been created and updated by surgeons using intraoperative ultrasound without ionizing radiation exposure or significant disruption to workflow [**5**].

Doppler ultrasonography can be utilized to design the surgical approach and evaluate the vascularity of the lesion. The direction and relative velocity of fluid along the probe's axis are determined by Doppler ultrasonography by using the Doppler effect, which is an observed frequency shift that occurs when a US wave is reflected back to the transducer from moving particles [**6**]. Based on this broad idea, there exist variations of Doppler imaging. Based on the recorded Doppler shift magnitude, color Doppler imaging overlays the 2D US image with a colored representation of the flow direction, either toward or away from the probe, in a specific area of the frame [**7**]. Color Doppler is highly angle dependent; no flow or Doppler shift will be seen at any place when the flow is perpendicular to the US waves. As a result, on a color Doppler, an angle change and a velocity change could seem identically [Figure 1] **[8].**

Additionally, color Doppler exhibits aliasing, an artifact caused by transducer pulse rate limits that causes portions of flow to be depicted with wrong magnitude or direction. Lastly, noise can greatly affect color Doppler imaging and overpower the flow signal [**9**]. As an alternative to color Doppler imaging, power doppler was developed. Power doppler depends on the strength of the Doppler shift signal rather than the size of the shift. Power Doppler features reduced noise, less angle dependence, better resolution for small vessels, and less aliasing when compared to color Doppler [**10**].

On the other hand, Power Doppler forfeits flow velocity and direction information. Additionally, blood vessel borders tend to be apparent to the power Doppler signal; hence, on power Doppler imaging, blood vessels appear larger than they do on MRA [**11**]. Furthermore, tiny vessels of little significance may be visualized due to the great sensitivity of the power Doppler, which would reduce its intraoperative utility. Lastly, compared to other imaging modalities, power Doppler has poor resolution overall and is limited by operator dependency and motion sensitivity errors [**12**].

ULTRASOUND PHYSICS

A piezoelectric transducer is used in diagnostic ultrasound to transform electrical signals into sound waves at frequencies higher than human hearing (1–20 MHz). Depending on the wavelength, frequency, and intrinsic physical acoustic properties of the tissue, these acoustic pressure waves are either absorbed, scattered, or reflected once they enter the tissue. The same piezoelectric transducer detects the sound waves that are reflected as echoes and transforms them into an electrical signal [**14**]. The tissues acoustic impedance (Z), which is derived from the product of the tissues density (ρ) and the sound velocity (c), which is related to the tissue's elasticity ($Z = \rho c$), is what determines the propagation of ultrasound. At tissue interfaces when there is a shift in acoustic impedance, sound waves are reflected. The magnitude of the reflected signal, which is related to an interface's acoustic gradient, determines how echogenic structures are on the US [**15**]. For example, the homogenous, low density, low acoustic impedance ventricles filled with CSF are hypoechoic, whereas the choroid plexus is hyperechoic and has a strong acoustic gradient with the surrounding brain. Diffuse reflectors, which are smaller interfaces that produce most of the echoes in the body, are what provide different tissues' distinctive speckled echotextures on ultrasound [**1**].

Acoustic energy is mostly dampened by absorption as heat and refraction in addition to reflection. Better resolution is achieved at the expense of more attenuation with higher US frequencies. As a consequence, low-frequency, lower resolution probes are better for seeing deeper structures and offering a wider field of view, and high-frequency probes are optimal for precise imaging of superficial structures [**16**].

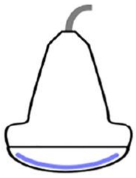
***Optimizing image quality*:** For maximum accuracy in any image-guided process, ideal image quality is crucial. Subpar image quality was linked to a far worse functional result following surgery. In doppler US, image quality varies greatly and is operator-dependent. Contrary to CT/MRI, which provides three-dimensional imaging of the entire head, intraoperative doppler ultrasonography can only view a small area from the craniotomy [**17**]. Depending on the location of the craniotomy, the type of probe, and the orientation of the probe, US can produce an endless number of different brain images. The novel perspective and tomographic depiction may be confusing to those who are not familiar with them. This learning curve is made steeper by the multitude of distinct probes, settings, and possible artifacts [**1**]. Obtaining high-quality images is possible with thorough planning and a consistent methodology. It is helpful to perform US sweeps in two orthogonal planes that are roughly equivalent to traditional anatomical planes in order to promote a methodical approach and make comparisons between US and MRI easier [**18**]. In order to enable generalization and comparability across various operators and units, standardization is also required. Building on this, an evaluation of the US's involvement in glioblastoma resection was carried out in the UK-based Functional and Ultrasound-Guided Resection of Glioblastoma (FUTURE-GB) randomized controlled study. This trial assesses the effect of resection guided by diffusion tensor imaging and US on deterioration-free survival. **[Figure 2] [19].**

Probe choice: There are several types of probes, and each has unique advantages and disadvantages. Small footprint probes are typically preferred for intraoperative use since they may be accommodated by the craniotomy. Transducers come in three primary varieties: sector array, curved, and linear. In the past, only big craniotomies could accommodate linear and curved transducers due to their huge footprints [**20**]. A low-frequency, small-footprint probe that produces a huge trapezoid field of view of the brain through a small craniotomy window is called a phased array, and it is one of the most commonly utilized types of sector array transducers. Phased array probes, sadly, have poor resolution and are especially prone to picture degradation. Better resolution linear and curved array probes with reduced footprints have recently been available **[Table 1] [21].** In a series that contrasted intraoperative Magnetic Resonance Imaging (iMRI) with a small footprint linear probe and a conventional phased array probe, linear ultrasound's sensitivity for tumor residual (79%) was significantly higher than that of iMRI (83%) and nearly equal to that of the phased array probe (21%).With an increased extent of resection (EOR) in 75% of cases, linear probes also show significantly superior imaging of vascularity and residual detection when compared to phased array probes [**22**].

Artifacts: Imaging artifacts may lead to over-aggressive resection of normal brain that has been mistakenly diagnosed as a tumor, or they may result in the missing diagnosis of residual illness. Acoustic shadowing (AS) and posterior wall acoustic enhancement (PAE) are the most commonly observed artifacts [**23**]. Acoustic shadowing happens close to interfaces where all US is reflected or absorbed; these are often seen near structures that strongly absorb sound energy, such as the brain-skull interface, or when there is a noticeable acoustic gradient. AS is frequently caused by gas bubbles in the surgical site or trapped in the sheathed ultrasonography probe. **[Figure 3][24].**

Ring-down artifact, which happens when an ultrasonic pulse comes into contact with tiny fluid collections caught between multiple gas bubbles, can also be caused by gas bubbles. The confined fluid resonates and sends a continuous signal back to the transducer, creating an artifact shadow that resembles an echogenic "step-ladder." Because hemostatic material can contain many gas locules, it is particularly known to be a cause of ring-down artifact [**1**]. Under fluid-containing homogeneous structures, such as cysts and fluid-filled resection voids, there is an auditory amplification of the posterior wall. Since fluid attenuates US less than solid tissue, stronger sound beams with larger amplitude and echogenicity are produced deep into the fluid. It can be challenging to distinguish between PAE and a persistent echogenic tumor at the bottom of a resection cavity [**25**]. Since PAE frequently has a linear morphology and occurs parallel to the US beam, detection can be aided by carefully evaluating variations in the PAE's appearance and adjusting the US probe. Reducing PAE can also be achieved by angularly positioning the probe on nearby preserved cortex that is angulated toward the resection's floor [**26**]. The surgical field can also seem different due to edema, contusion, and coagulated blood. Blood and contusions provide special challenges because they resemble residual illness characteristics and seem echogenic. When distinguishing residuum from other surgically linked alterations, intravitreal linear transducers perform better [**27**]. A lesion will be visible on all pictures, but artifacts such PAE and surgical alterations will have formed over the surgery period. Careful connection with the preoperative navigation MRI and previous earlier US scans is crucial [**28**].

Table (1): Summary table of different types of ultrasound probes and potential use cases **[1].**

Appearance	Type	Depth (cm)	Description and use cases
	Linear array 7-15 MHz	2-7	Very high spatial resolution but limited penetration. Creates a rectangular field of view with less artifact compared to convex arrays. Typically large foot-print so limited to large craniotomies.
	Convex array 2-10 MHz	10-30	High spatial resolution with good penetration. Fan shaped large field of view. Large foot-print so limited to large craniotomies.
	Micro-convex 4-13 MHz	6-10	High spatial resolution with good penetration. Fan shaped large field of view. Smaller foot-print so more adaptable and usable in smaller craniotomies with potential for intracavitary use depending on resection size.
	Sector array 4-10 MHz	4-8	Small foot-print. Produces trapezoid image allowing wide field of view from a small craniotomy. Resolution lower at depth. Can be used for burr-hole guided surgery for instance for VP shunt placement.
	Matrix phased array 1-8 MHz	5-20	Type of sector array and often used in neurosurgery. Allows direct easy acquisition of a pyramidal 3D US image allowing volumetric reconstruction in any axis and facilitating visualization of adjacent structures. Produces relatively large field of view but resolution and contrast between different structures is poorer versus linear and convex array probes.
	Small linear "Hockey-stick" 6-15 MHz	2-5	Small foot-print, very high resolution but limited penetration. Can be placed directly into the resection cavity for high resolution assessment of superficial residual disease at the resection margin.

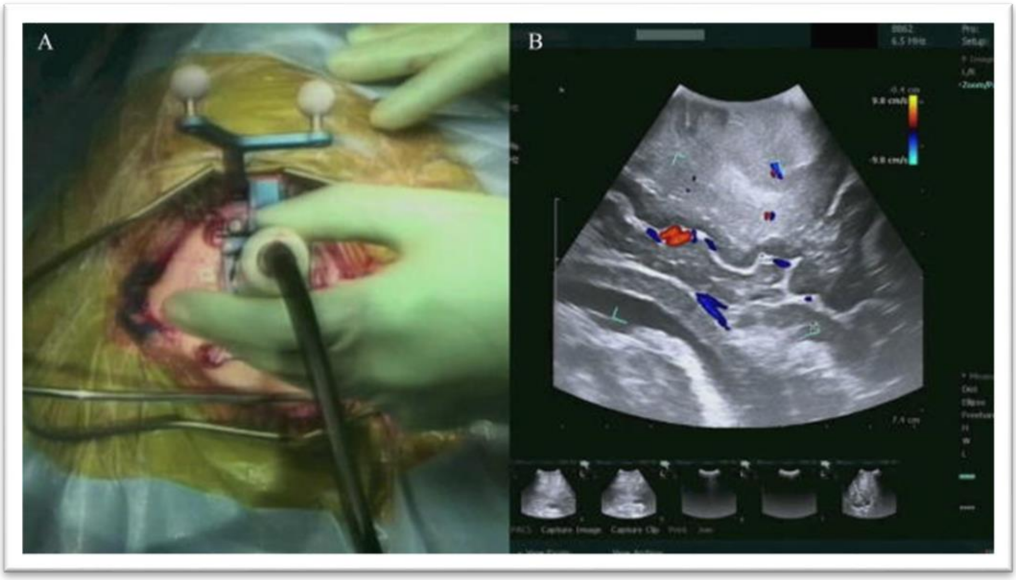


Figure (1): (A) Intraoperative 2D tracked ultrasound image acquisition in the operating room. Fiducials for intraoperative optical neuron avigation are visible rigidly attached to the ultrasound probe. (B) 2D ultrasound image with superimposed color Doppler imaging [**13].**

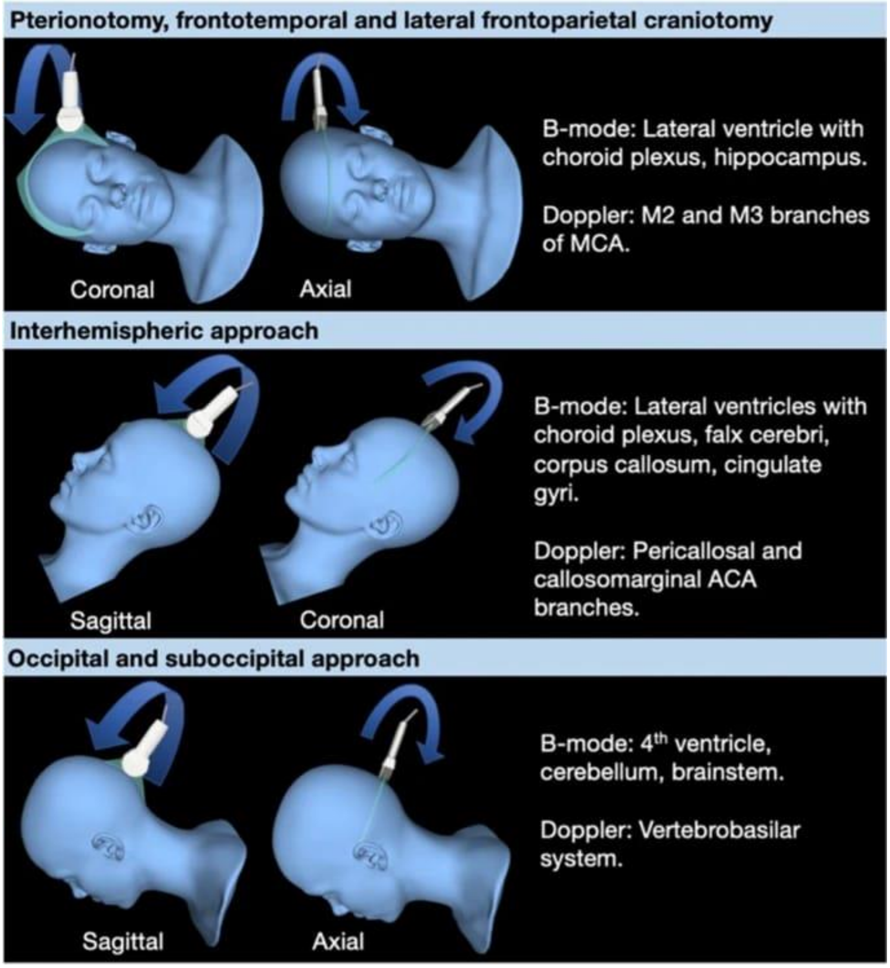


Figure (2): Recommended orthogonal ultrasound fans for different craniotomies with expected anatomical and vascular landmarks. Model of orthogonal ultrasound sweeps for common craniotomy sites. Probe positioned to achieve views that approximate to standard anatomical planes on CT/MRI. Patient positioned to ensure the craniotomy is as horizontal as possible to allow retention of fluid in the resection cavity for optimal ultrasound coupling [**1].**

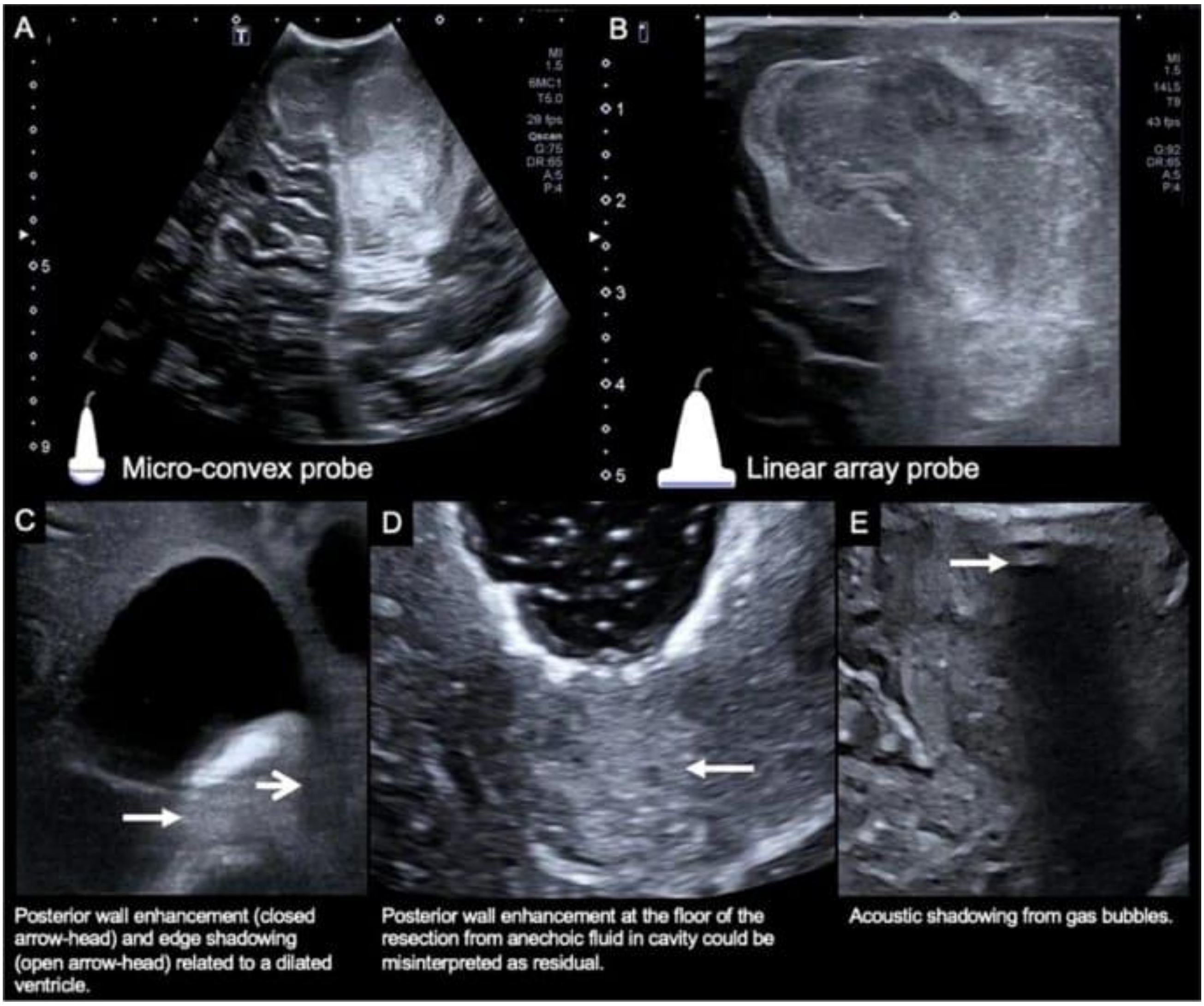


Figure (3): Example of ultrasound images from different probes and common ultrasound artifacts. Microconvex (**A**) and linear (**B**) US probe images of a medulloblastoma in the left cerebellum. Note the large field of view permitted by the microconvex probe (A) but the relatively poor resolution compared to the small field of view image arising from the linear probe (B).
Posterior wall enhancement (closed arrowhead) and edge shadowing (open arrowhead) related to the frontal horn of the lateral ventricle (**C**). Posterior wall enhancement at the floor of a resection cavity secondary to anechoic fluid in the resection cavity could be misinterpreted as residual disease (**D**). Acoustic shadowing from gas bubbles obscures the central field of view (**E**) [**28].**

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