Comparison between ultrasound-guided and anatomical landmark-guided cannulation of the internal jugular vein in pediatrics.

Thesis

Submitted by

Nancy Nabil Hamza Salem

M.B.B.ch.

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Supervised by

Prof. Dr. Laila Halim Doss

Professor of Anesthesiology and Intensive Care

Faculty of Medicine

Cairo University

Prof. Dr. Maha Gamil Hanna

Professor of Anesthesiology and Intensive Care

Faculty of Medicine

Cairo University

Dr. Passaint Mohammed Fahim

Lecturer of Anesthesiology and Intensive Care

Faculty of Medicine

Cairo University

Cairo University

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- ***Nancy Nabil Hamza***

Abstract

Successful CVC insertion requires minimizing placement complications and procedure time. The use of ultrasound to guide the insertion allows direct visualization of target vessel. Ultrasonographic localization of the internal jugular vein in pediatrics did not increase the success rate of CVC insertion, but decreased the number of total skin punctures, time needed for insertion, and incidence of carotid artery puncture, even it was not a significant difference.

Key word: CVC- Jugular vein- US- pediatrics

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List of abbreviations

HZ	hertz
kHZ	kilohertz
MHZ	megahertz
Z	Acoustic impendance
dB	**decibell**
SVC	Superior vena cava
CVC	Central venous catheter**.**
TPN	Total parenteral nutrition
PA	Pulmonary artery.
IJV	Internal jugular vein
SV	Subclavian vein
FV	Femoral vein
SAX	Short axis
LAX	Long axis
CA	Carotid artery
UAI	Ultrasonography-assisted insertion
US	Ultrasound
LM	Land mark
NICE	The National Institute for Clinical Excellence

Introduction

Central venous catheters (CVC's) are used to provide secure access to the central circulation in patients who require invasive hemodynamic monitoring, inotropic support, temporary transvenous cardiac pacing, or in whom adequate peripheral venous access is unobtainable. [1]

Successful CVC insertion requires minimizing placement complications such as multiple puncture attempts, arterial puncture, nerve puncture, pneumothorax, or hematoma and minimizing procedure time. [2]

We are now entering an era in which the scope of ultrasound is expanding. The use of ultrasound (US) to guide CVC placement allows direct visualization of target vessel, assessment of anatomic variations and vein thrombosis. [3]

Ultrasound-guided central venous catheter insertion is established as a gold standard for the placement of internal jugular vein lines and its use has extended to other vascular access sites. [4, 5] Ultrasound has been shown to increase success rates and decrease the number of procedural complications in adults. [6, 7]

The use of ultrasound for central line placement allows for faster line placement and a lower rate of arterial catheterization when compared with line placement using the landmark technique. [8]

In the pediatric population, some studies report high success rates and superiority compared to the standard landmark technique when using ultrasound for central line placement. [9, 10]

PHYSICAL PRINCIPLES OF ULTRASOUND

Discovery of Ultrasound:

Many species including bats use ultrasound to navigate flight and to locate food sources. The first detailed experiments that indicated that non-audible sound might exist were performed on bats by Lazzaro Spallanzani (1729–1799) an Italian priest and physiologist (11). Spallanzani concluded that 'The ear of the bat serves more efficiently (than the eye) for seeing, or at least for measuring distances, a matter of scientific heresy in the 1790s. 'Spallanzani's bat problem', as it was termed, remained a scientific mystery until 1938, when finally the young Harvard students, Donald R. Griffin and Robert Galambos used a sonic detector to record directional ultrasound noises being emitted by bats in navigating flight (12).

The application of directional sound reflections being used to detect objects and measure distances was initially developed for nautical purposes. After the sinking of the Titanic, devices using active echolocation were patented in 1912, with the first sonar (sound navigation and ranging) apparatus being built in 1914, capable of detecting an iceberg 2 miles away. The threat of German submarines to Allied shipping in World War I provided a pressing impetus to the development of ultrasound technology (13). Between the wars, ultrasound techniques were applied to detect flaws in metal (in particular in ships and aircraft) using machines called reflectoscopes or flaw detectors (14). These military and industrial applications of ultrasound were to lead to the development of medical diagnostic ultrasound.

The use of ultrasound as a medical diagnostic tool began in 1942 when Karl Dussik, a neurologist at the University of Vienna, attempted to locate brain tumours and the cerebral ventricles by measuring the transmission of ultrasound beams through the head (15).

Professor Ian Donald of Glasgow first developed a two-dimensional scanner and then an automatic scanner in 1960, made the first ante-partum diagnosis of placenta previa using ultrasound, developed the method for measuring the biparietal diameter of the fetal head in 1962 and was the first to utilize the full bladder to allow the detection of very early pregnancy of about 6–7 weeks gestation in 1963 [16].

Production of Ultrasound

Ultrasound vibrations (or waves) are produced by a very small but rapid push–pull action of a probe (transducer) held against a material (medium) such as tissue. Virtually all types of vibration are referred to as acoustic, whereas those of too high a pitch for the human ear to detect are also called ultrasonic. Vibrations at rates of less than about 20 000 push–pull cycles/s are audible sound, above this the term ultrasonic is employed. In medical ultrasound, vibrations in the range 20 000 to 50 000 000 cycles/s are used. The term frequency is employed rather than rate of vibration and the unit hertz (Hz) rather than cycles/s. We therefore use frequencies in the range 20 kilohertz (20 kHz) to 50 megahertz (50 MHz). In medical ultrasound the source is a piezoelectric crystal. Conversely, when ultrasound waves strike a piezoelectric crystal causing it to vibrate, electrical voltages are generated across the crystal, hence the ultrasound is said to be detected. The hand-held devices containing piezoelectric crystals and probably some electronics are called transducers since they convert electrical to mechanical energy and vice versa. They are fragile and expensive, about the same price as a motor car. The basic data for most ultrasound techniques is obtained by detecting the echoes which are generated by reflection or scattering of the transmitted ultrasound at changes in tissue structure within the body. [17]

The push–pull action of the transducer causes regions of compression and rarefaction to pass out from the transducer face into the tissue. These regions have increased or decreased tissue density. A waveform can be drawn to represent these regions of increased and decreased pressure and we say that the transducer has generated an ultrasound wave. The distance between equivalent points on the waveform is called the wavelength and the maximum pressure fluctuation is the wave amplitude. If ultrasound is generated by a transducer with a flat face, regions of equal compression or rarefaction will lie in planes as the vibration passes through the medium. Plane waves or wavefronts are said to have been generated. Similarly if the transducer face is convex or concave the wavefront will be convex or concave. The latter can be used to provide a focused region at a specified distance from the transducer face. In tissue if we could look closely at a particular point, we would see that the tissue is oscillating rapidly back and forward about its rest position. As noted above, the number of oscillations per second is the frequency of the wave. The speed with which the wave passes through the tissue is very high close to 1540 m/s for most soft tissue. The speed of sound, c, is simply related to the frequency, f, and the wavelength, λ, by the formula:

$$c = f\lambda$$

In the acoustic waves just described the oscillations of the particles of the medium are in the same direction as the wave travel. (17)

This type of wave is called a longitudinal wave or compressional wave since it gives rise to regions of increased and decreased pressure. In another type of wave the oscillations are perpendicular to the direction of wave travel, and they are called transverse or shear waves. At MHz frequencies the latter are attenuated rapidly in tissues and fluids and hence are not encountered at present in diagnostic ultrasound. The speed of ultrasound in soft tissue depends on its rigidity and density (Table 1), the more rigid a material, the higher the speed. From the table we can see that the speed in soft tissues is fairly closely clustered around an average value of 1540 m/s.Virtually all diagnostic instruments measure the time of echo return after the instant of pulsed ultrasound transmission and then use the speed in tissue to convert this time into the tissue depth of the reflecting structure. For a mixture of soft tissues along the pulse path, an accurate measure of depth is obtained by the assumption of an average speed of 1540 m/s for conversion of time into depth. The high speed in bone can cause severe problems as will be seen later when refraction of the ultrasound is considered. Unless bone is thin it is best avoided during ultrasound examinations. The speed of ultrasound in soft tissue is independent of frequency over the diagnostic range, i.e. 1 to 50 MHz. The very high value of the speed of sound in tissue means that echo data can be collected very rapidly. (17)

(Table 1) Speed of ultrasound and acoustic impedance (17)

Material	**Speed (m/s)**	**Acoustic impedance(g/cm2 s)**
Water (20°C)	1480	1.48 × 105
Blood	1570	1.61 × 105
Bone	3500	7.80 × 105
Fat	1450	1.38 × 105
Muscle	1580	1.70 × 105
Air	330	0.0004 × 105
Soft tissue (average)	1540	1.63 × 105

Reflection

Ultrasound is reflected when it strikes the boundary between two media where there is a change in density or compressibility or both.
To be more exact, reflection occurs where there is a difference of acoustic impedance (Z) between the media (fig.1). The impedance is a measure of how readily tissue particles move under the influence of the wave pressure. The acoustic impedance of a medium equals the ratio of the pressure acting on the particles of the medium divided by the resulting velocity of motion of the particles. Therefore for tissues of different impedance a passing wave of particular pressure produces different velocities in each tissue. The velocity of particle motion about the rest position is not the same as the velocity (speed) of the ultrasound waveform through the medium. For a wave in which the peaks and troughs of pressure lie in flat planes, plane waves, the acoustic impedance of a medium is equal to the density (ρ) times the speed of sound in the medium, i.e. $Z = \rho c$. It is not surprising that reflection depends on quantities such as density and speed of sound, since the latter depends on the rigidity of the medium. In practice when imaging, echo size is often related roughly to the change in acoustic impedance at tissue boundaries. Note that it is the change that is important – it does not matter whether there is an increase or decrease in impedance. The large changes in acoustic impedance at bone/soft tissue and gas/soft tissue boundaries are problematic since the transmitted pulse is then greatly reduced or even totally blocked in the case of gas by reflection at the boundary.

The size of the echo in an image (i.e. the shade of grey) relates to the change in acoustic impedance at the interface producing it. Shades of grey are therefore related to the properties of tissues though signal processing in the scanner also plays an important part. The unit of acoustic impedance is the Rayl, where 1 Rayl

= 1 kg/m^2s (units are often named after individuals). The higher the density or stiffness of a material, the higher is its acoustic impedance.

It is instructive to consider the simple case of reflection of an incident ultrasound wave at a flat boundary between two media of impedances ρ1c1 and ρ2c2. The magnitude of the echo amplitude is calculated using:

Reflected amplitude = Incident amplitude × $(\rho_1 c_1 - \rho_2 c_2)/(\rho_1 c_1 + \rho_2 c_2)$ Reflection of ultrasound at a smooth surface is similar to light reflecting at a mirror and is sometimes referred to as specular reflection. The angle of incidence, i, is equal to the angle of reflection, r. [17]

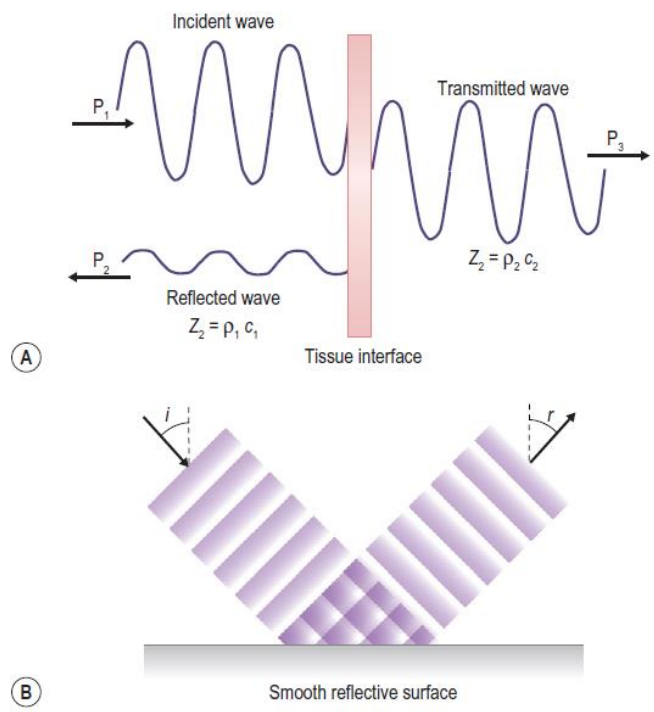

Figure 1 Reflection. **A**: Reflection at change in acoustic impedance between two media. **B**: Reflection at a smooth interface. [17]