TRANSESOPHAGEAL DOPPLER DEVICES: A TECHNICAL REVIEW

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ABSTRACT. Monitoring of aortic blood flow, conducting large portions of the cardiac output (CO), allows conclusions on the global hemodynamic status of patients. For this purpose, transesophageal Doppler (TED) devices have been developed, which interrogate the descending aorta and calculate aortic blood flow velocity using the Doppler principle. The recorded velocity–time curve can be used to estimate CO as well other advanced hemodynamic parameters such as preload, afterload and myocardial contractility. Clinical studies in perioperative patients have demonstrated a reduced postoperative morbidity and shorter length of hospital stay when TED is used to guide fluid management. However, several assumptions are needed to translate the measured Doppler frequency shift to hemodynamic variables and discrepancies between the assumed and the actual condition may introduce a considerable risk for erroneous calculations. A correct interpretation of the displayed parameters requires profound knowledge on the technical basis of this method as well as its technical limitations. Our review focuses on these technical aspects which the clinician should be familiar with to allow proper use of TED monitoring devices.

KEY WORDS. Doppler, ultrasound, cardiac output, stroke volume, hemodynamic monitoring.

INTRODUCTION

Diagnostic ultrasound is increasingly used in anesthesiology and intensive care medicine. On the one hand, B-mode ultrasound allows clinicians to identify anatomical structures and can thus facilitate vascular puncture [1, 2] or regional anesthesia [3, 4]. On the other hand, Doppler ultrasound is used to assess blood flow within vessels to monitor regional perfusion of different organs such as the brain, kidneys or uterus [5–9]. Measurements of blood flow in the aorta, conducting large portions of the cardiac output (CO), allow global hemodynamic monitoring. For this purpose, Transesophageal Doppler (TED) devices have been developed, which interrogate the descending aortic blood flow and permit estimations of CO, preload, afterload and myocardial contractility. In general, TED might be a less invasive alternative to other monitoring techniques such as thermodilution using a pulmonary artery catheter. A major advantage of TED is that it allows extended monitoring and facilitates hemodynamic optimization in patients in whom advanced

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hemodynamic parameters such as CO would otherwise not be monitored. Compared to conventional clinical assessment, fluid management guided by TED to optimize stroke volume has been shown to improve patient outcome and to reduce the length of hospital stay [10–19], suggesting that this technique can be a valuable supplement to current standard hemodynamic monitoring.

Early TED devices were complex to operate which prevented a widespread use. More recently, modern, user-friendly TED devices have been developed making this technique suitable for routine clinical use. The growing interest in TED monitoring is reflected by the publication of recent review articles, which mainly focus on clinical aspects of the technology [20, 21]. However, correct interpretations of the displayed parameters require a thorough understanding of the technical basis and the assumptions which are required to translate the measured Doppler frequency shift into hemodynamic parameters. Therefore, this non-systematic review article focuses on technical aspects and limitations which the clinician needs to consider to allow rational use of the technology.

THE DOPPLER EFFECT

An apparent change in the frequency of a wave noted by an observer moving relative to the source of the wave (Figure 1) was first proposed by the Austrian physicist Christian Andreas Doppler (1803–1853) in 1842 at a meeting of the Royal Bohemian Society of Sciences in Prague [22]. Doppler compared this effect, which has subsequently been termed the “Doppler Effect”, with a ship which crosses waves more frequently when heading against the waves than when moving in the same direction as the waves. The Dutch scientist Buys Ballot (1817–1890), actually a critic of Doppler aiming to disprove his theory, supplied first empiric evidence of the Doppler Effect on the railway between Amsterdam and Utrecht (the Netherlands) [23, 24]. Since accurate measurement of wavelengths was not possible at that time, musicians with a skilled sense of hearing were positioned at the rails and asked to rate the pitch of a brass instrument played on the train and vice versa. Indeed, the perceived pitch appeared higher as the actually played tone whenever the instrument and the musician approached and lower as they departed from each other [23, 24].

Waves reflected by moving objects also undergo a Doppler frequency shift. The total Doppler effect results from motion of the wave emitter, the reflecting object and the receiver. The discrepancy between actual and noted frequency, i.e. the frequency shift, is directly pro-

Fig. 1. The Doppler effect describes an apparent change in the frequency of a wave when the emitter of the wave and the receiver move relative to each other. The perceived frequency is higher than the actually emitted frequency when the emitter (E) approaches the receiver (R₁) and lower when the emitter moves away from the receiver (R₂). The frequency shift is proportional to the relative velocity between emitter and receiver.

portional to the relative velocity between emitter and receiver. Therefore, if the emitter or receiver are stationary or move with a known speed and direction, the velocity and direction of the moving object can be determined. Nowadays, this principle is widely used in radar systems, such as in air traffic control radar, weather radars, and police speed guns. In medicine, Doppler sonography is used by applying the Doppler principle to ultrasound technology.

TRANSESOPHAGEAL DOPPLER SONOGRAPHY

Enhancing ultrasound with measurements of Doppler frequency shifts allows us to determine blood flow velocity. Herein, ultrasound emitted by a transducer is reflected by moving red blood cells (Figure 2) [25]. Basically, two types of Doppler Sonography can be distinguished: in continuous wave Doppler (CWD), two adjacent transducers couple continuous transmission of ultrasound with continuous reception of reflected waves, whereas Pulsed wave Doppler (PWD) only uses one transducer which intermittently transmits and receives [26]. The physical characteristics of the emitted ultrasound waves depend on their frequencies, which usually vary between 1 and 10 MHz for medical purposes. Tissue penetration depth is inversely related to the frequency whereas the intensity of the reflected sound is proportional to the fourth power of the frequency [26, 27]. Therefore, the “optimal” frequency for each application depends on the depth of the structure to be examined and represents the best compromise between signal strength and tissue penetration.

The blood flow velocity (v) is directly proportional to the frequency shift (Δf) and can be calculated by the standard Doppler equation [26, 28, 29]:

Fig. 2. Doppler ultrasound transducers directed towards a blood vessel at different angles. When the blood flow is directed towards the transducer (A), the frequency of the reflected ultrasound waves increases as compared to the transmitted frequency. In contrast, blood flow away from the transducer (B) is associated with a decrease in reflected ultrasound frequency. Ultrasound insonated at 90° (C) does not undergo any frequency shift, because erythrocytes neither move towards nor away from the transducer. When the vessel is insonated at very shallow angles (D), a major proportion of the waves is reflected by the vessel wall and does not reach the lumen.

$$v = \frac{\Delta f \times c}{2f_T \times \cos \theta}$$

where c is the velocity of the ultrasound waves in body tissue (average $1,540 \text{ m s}^{-1}$ in soft tissue [28]) and f_T the emitted frequency. The cosine of the angle between Doppler beam and blood flow ($\cos \theta$) is a correction factor needed to adjust for the angle of insonation. Maximal Doppler shift is obtained when the erythrocytes move directly towards or away from the receiver in a straight line (0°) whereas ultrasound insonated at 90° does not undergo any Doppler shift because no component of the flow is parallel to the ultrasound beam (Figure 2). The cosine changes slowly at shallow angles due to the non-linear character of the cosine function but rapidly changes at increasing angles. Therefore deviations of the actual from the assumed angle of insonation will result only in a moderate error in flow velocity calculation with shallow angles but becomes unacceptably high with angles exceeding $\sim 60^\circ$ (Figure 3). Thus, the angle of insonation should ideally be as close to 0° as possible. However, as the ultrasound beam passes through the wall of the vessel, a certain proportion is reflected and does not reach the lumen. The extent of reflection is higher at shallow

angles, technically precluding very shallow angles due to nearly complete wave reflection (Figure 2). Thus, insonation angles between 30° and 60° are considered useful [26].

Franklin et al. are credited for performing first prototypical blood flow measurements with Doppler sonography in 1961 by clamping two transducers on the aorta of a dog [30], however, in fact Satomura had already published an article in a Japanese journal about Doppler flow measurements in humans 2 years earlier [31]. Shortly after these pioneer articles, non-invasive transcutaneous transducers were developed for clinical use [32, 33]. In subsequent studies, the ultrasound beam was directed from an intercostal space or suprasternal notch towards the aortic arch or ascending aorta, allowing non-invasive determinations of aortic blood flow in humans [34, 35]. Continuous monitoring was, however, complicated because it was hardly possible to fix an external transducer in place and to avoid changes in insonation angle or loss of the signal. In contrast, esophageal placement of the probe, first described by Side and Gosling [36], allows to keep it in place in close proximity to the descending aorta. The esophagus and descending aorta run almost parallel at the mid-thoracic level. Therefore, the insonation angle is

Fig. 3. Percent error in cosine of the angle of insonation (θ) resulting from 1° (A) and 5° (B) deviation in the actual from the assumed angle. Note that the error is negligible or moderate at shallow angles but markedly increases at higher angles.

approximately the same as that between the probe and the transducer if the transducer is mounted to the probe in a fixed angle [37–39], allowing interrogation of descending aortic blood flow with a stable, known angle of insonation. This technique of transesophageal Doppler measurements of aortic blood flow velocity has subsequently been further refined by other investigators [37, 39–45], and several commercial devices have been developed. Most of these devices, however, are no longer marketed as they turned out to be complex and not user friendly, became technically outdated or were bought up and abandoned by competitors. Currently, only two devices are broadly used, which employ 4 MHz CWD and an angle of insonation of 45° (CardioQ, Deltex Medical Ltd., Chichester, UK) or 5 MHz PWD with an angle of insonation of 60° (HemoSonic 100, Arrow International, Reading, PA, USA).

TRANSLATING AORTIC BLOOD FLOW VELOCITY INTO HEMODYNAMIC PARAMETERS

As erythrocytes pass the ultrasound beam in the descending aorta, blood flow velocity is calculated based on the Doppler equation. The systolic portion of the velocity–time envelope is typically triangular and the area under this curve ($\text{cm s}^{-1} \text{s}$) represents the stroke distance (cm), i.e. the distance that the blood column has moved forward in the aorta during systole (Figure 4). Stroke distance is proportional to stroke volume but the exact formula for the determination of stroke volume from stroke distance varies according to the device used. The

HemoSonic 100 device (Arrow International, Reading, PA, USA) provides real-time measurements of the aortic diameter via an integrated 10 MHz M-mode ultrasound probe orientated at a 90° angle to the central axis of the probe. Assuming a circular shape of the aorta, the aortic diameter is used to calculate aortic cross sectional area. By multiplying stroke distance (cm) with aortic cross sectional area (cm^2), the descending aortic stroke volume (cm^3) can be determined (Figure 4). Blood flow in the descending aorta is only a fraction of total CO, therefore a constant proportion of blood flow between the descending aorta (~70%) and the coronary and brachiocephalic arteries (~30%) is assumed to estimate systemic stroke volume. Instead of measuring the aortic diameter, the CardioQ device (Deltex Medical Ltd., Chichester, UK) estimates CO from a nomogram based on the patient's age, height and weight [46].

The aortic blood flow velocity–time envelope also allows the calculation of other advanced hemodynamic parameters which can be useful to estimate preload, afterload and myocardial contractility. The base of the triangular systolic portion of the plot represents the systolic ejection time (Fig. 4). This ejection time, frequently referred to as “flow time”, depends on the heart rate. In analogy to correcting the QT interval of an ECG using Bazett's equation [47], flow time is corrected to one cardiac cycle per second (“flow time corrected”, FTc) by dividing it by the square root of the cycle time. Acceleration of blood in the descending aorta during early systole corresponds to the upslope of the velocity–time plot, from which the mean acceleration can be determined. The peak of the triangle marks the peak blood

Fig. 4. Velocity–time plot and calculation of descending aortic blood flow. The systolic portion of the velocity–time plot is typically triangular and its base represents the flow time (= systolic ejection time). The upslope of the graph depicts the acceleration of blood and the peak of the waveform shows the peak velocity in the descending aorta. The area under the systolic portion of the curve (AUC) represents the stroke distance, i.e. the distance that the blood column has moved forward in the aorta during systole. The descending aortic stroke volume can be determined by multiplying the stroke distance with the aortic cross sectional area.

Fig. 5. Velocity–time waveforms of aortic blood flow under three hemodynamic conditions, which are all associated with a reduction in cardiac output but require different therapeutic approaches. Note that the waveform itself as well as the reported hemodynamic parameters can facilitate the differential diagnosis. SV stroke volume, FTc flow time corrected, PV peak velocity, MA mean acceleration, ↑ and ↓ direction of change.

flow velocity during systole, followed by a down-slope which depicts deceleration of flow during later systole (Figure 4).

Reference values of these parameters have only poorly been established and changes in the parameters relative to their baseline value may be more informative than their

absolute value. Since systole takes approximately one-third of the entire cardiac cycle, an FTc of about 330 milliseconds (ms) is generally considered normal [48], however, a shorter FTc of about 290 ms may be normal in young adults [49]. The peak velocity decreases with increasing age from ~ 100 to 130 cm s^{-1} at an age of 20 years to about $60\text{--}70 \text{ cm s}^{-1}$ at 70 years [49, 50]. Ascending aortic blood flow measurements suggest that mean acceleration also decreases with increasing age [49].

The FTc primarily responds to changes in pre and afterload, while peak velocity and mean acceleration are predominantly considered markers of left ventricular contractility. However, it is important to realize that no single parameter is specific for preload, afterload or contractility and that changes in one parameter are in vivo accompanied by compensatory changes of other parameters. Only the combination of the different parameters and their response to dynamic cardiovascular events or therapeutic interventions can give a comprehensive picture of the hemodynamic situation. Typical responses of the TED key parameters stroke volume, FTc, peak velocity and mean acceleration to varying hemodynamic conditions have been reviewed previously [20, 21, 51] and are summarized in Figure 5.

CLINICAL USE

Technically, the insertion of a TED probe is similar to the placement of a gastric tube [20]. Insertion depth is typically between 35 and 45 cm for adults, depending on the route of insertion (oral vs. nasal route). The probe is rotated until the ultrasound beam aligns with the descending aorta as confirmed by visualization of the typical aortic waveform and a characteristic pulsatile sound pattern. Optimal probe focus is accomplished by slight manipulations of the probe until the largest and sharpest possible waveform is found. The integrated M-mode of the HemoSonic 100 device also allows direct visualisation of the aortic wall. Contraindications for the insertion of TED probes include any pathology which predisposes the patient to an increased risk of injury at the insertion site or esophagus, such as malformations, strictures, tumors, varices, esophagitis, or severe bleeding disorders [20].

TED-derived measurements of CO have been compared to reference methods, most often pulmonary artery catheter thermodilution techniques [38, 43, 45, 52–92]. The pooled median bias in the studies reporting Bland–Altman analyses is moderate (0.37 l min^{-1}), however, with wide “95% limits of agreement”-values (median 4.2 l min^{-1}) [20]. This demonstrates that TED does not systematically over or underestimate CO, however,

individual CO measurements obtained through TED may differ considerably from CO values derived by the thermodilution technique. The observed discrepancies between the absolute values measured with both techniques may in part be explained by the following factors. First, pulmonary artery catheter thermodilution techniques assess right ventricular output while TED measures aortic blood flow velocity, which is a function of left ventricular output. Right and left ventricular output are not necessarily identical, i.e. due to draining of venous blood from Thebesian and pulmonary veins also directly into the left heart. Second, although the thermodilution technique is considered the clinical “gold standard”, it also has methodological limitations and does not necessarily reflect the “true” CO under all conditions [93, 94]. Third, some of the validation studies were performed under conditions in which basic assumptions of TED measurement concerning a constant diversion of blood flow were violated, e.g. by aortic cross clamping or epidural sympathectomy [63, 73, 82, 85].

In clinical practice exact values are usually less relevant than information about changes in CO. The median correlation reported between TED and reference methods is 0.80, suggesting that a high value obtained with one technique will likely be reflected by a high reading with the other technique and vice versa, and that the direction of changes in CO can be tracked by TED [20]. Indeed, numerous studies report that TED measurements accurately follow changes in CO over time [43, 54, 55, 60, 63, 74, 78, 80, 82, 83, 85].

In critically ill patients, TED can be used for advanced hemodynamic management. However, up to now data to support its usefulness in guiding inotropic or vasoactive therapy in such patients is limited and the potential of TED to replace more invasive techniques in critical care units needs to be further determined. In contrast, the role of TED in optimizing perioperative volume replacement has been well characterized [11–19]. TED allows assessment of fluid responsiveness by monitoring changes in stroke volume secondary to fluid challenges, allowing individual titration of fluids to maximize CO while avoiding excessive fluid loading.

Clinical studies comparing TED guided protocols to conventional approaches of volume replacement (guided by clinical assessment and/or central venous pressure) conclusively report beneficial effects in the Doppler-optimized groups, including a reduced risk of postoperative morbidity and a shorter length of hospital or ICU stay [11–19]. In addition to fluid optimization, other clinical applications such as early detection of hemodynamic changes associated with transurethral resection syndrome have also been reported [95].

TECHNICAL LIMITATIONS

Limitations in measurement of blood flow velocity

Transesophageal Doppler devices basically measure only two parameters: the Doppler frequency shift and the elapsed time. Blood flow velocity is calculated from the frequency shift of ultrasound reflected by red blood cells under the assumption that all erythrocytes travel in the same direction and at the same speed. The pulsatile blood flow in combination with a complex aortic geometry including curvatures in multiple planes, tapering, branches, and distensible walls results in a complex three-dimensional aortic flow pattern [96, 97]. Descending aortic blood flow is usually more or less laminar and shows a relative uniform velocity profile over the aortic cross section in healthy subjects [98]. However, the existence of skewed velocity profiles and rotational blood flow in the descending aorta has also been suggested [96]. Non-laminar flow might be expected in patients with pathology of the aorta or aortic valve, such as dissection, aneurysm, coarctation or valvular stenosis.

The angle of insonation must be known and remain constant for accurate velocity measurements. While this angle approximates the angle between the probe and the transducer in healthy persons, the above assumption may not hold true during surgical manipulations affecting the anatomic relationship between esophagus and aorta or in patients with altered thoracic anatomy, e.g. due to tumor mass, previous surgery or scoliosis.

Furthermore, the accuracy of velocity measurements principally depends on the chosen Doppler mode. CWD lacks depth resolution and therefore interrogates the blood flow in all vessels in the range of the ultrasound beam. Measurement of non-aortic blood flow, however, is negligible because there are no other major vessels in the range of the ultrasound beam when it is orientated from the oesophagus towards the descending aorta. PWD does allow axial resolution, enabling selective interrogation of aortic blood flow, however, at the cost of inaccuracy at high blood velocities [26, 27]. The maximum velocity which can be reliably measured depends on the pulse repetition frequency (PRF), which is for instance ~ 12.4 kHz for the HemoSonic 100 device. According to the Nyquist–Shannon sampling theorem, the maximum frequency shift that can be measured is one half of the PRF [26]. This maximum measurable frequency shift corresponds to a maximum detectable blood velocity of ~ 190 cm s⁻¹ when the frequency and angle of insonation of the HemoSonic 100 device are applied to the Doppler equation. Since this velocity considerably exceeds peak velocities normally observed in the aorta,

we conclude that PWD as well as CWD are both similarly suitable for transesophageal determinations of aortic blood flow velocities despite their theoretical limitations.

Limitations in the estimation of stroke volume and cardiac output

After the aortic blood flow velocity has been determined, other hemodynamic parameters can be calculated. FTc, mean acceleration and stroke distance are calculated directly from the velocity–time–function and can therefore be regarded as relatively “robust” parameters. In contrast, calculations of stroke volume and CO require additional assumptions or estimations, introducing a potential for erroneous calculations. In fact, the need for a volumetric translation of TED data has been questioned due to these additional limitations [43].

The first drawback of calculating stroke volume is that the aortic cross sectional area needs to be known or estimated. One possibility is to measure the aortic diameter by M-mode ultrasound (for instance HemoSonic 100 device, Arrow International, Reading, PA, USA). This is complicated because the aortic diameter is not constant as ejected blood distends the aortic wall during systole [99, 100]. Moreover, the formula used to calculate aortic cross sectional area assumes a circular shape ($A = \pi r^2$), however, the aorta is not a perfect circle. Neither is the aorta as cylindrical as depicted in Figure 4, rather it tapers from cranial to caudal giving it a slightly conical aspect. Another possibility is to circumvent these difficulties in measuring aortic diameter by using a nomogram to translate blood flow velocity to CO (for instance CardioQ, Deltex Medical Ltd., Chichester, UK) [46]. Nevertheless, a nomogram derived from average values in a population does not necessarily accurately predict individual values. Moreover, the nomogram does not account for changes in aortic diameter. Such changes may occur due to hemodynamic changes, e.g. blood loss or changes in blood pressure [100–102]. One study showed an inconsistent association between Doppler derived CO measurements and CO measured according to the Fick-principle in an experimental animal hemorrhage model [103]. The authors did not use the CardioQ nomogram but a calibration factor which was individually determined for each animal. In this study, TED often indicated an increase in CO in the face of ongoing blood loss, decreasing blood volume and arterial hypotension. The authors did not account for changes in the aortic diameter when calculating CO and conclude that especially these unaccounted changes in aortic diameter were responsible for the inconsistent and unpredictable values of CO they observed [103]. Users of TED should be aware that similar

limitations may likely apply to every other calibration factor or nomogram which does not consider changes in aortic diameter.

Calculation of the stroke volume assumes a constant diversion of blood flow between supra-aortic vessels and the descending aorta. However, this proportion is not constant. Changes in vascular tone, e.g. due to general anesthetics, vasodilatation (e.g. due to sympathetic blockade during neuraxial anesthesia) as well as vasoconstriction (e.g. compensatory due to blood loss or mechanically during aortic cross clamping) may affect the distribution of CO [73, 104–106]. It should also be noted that in patients with aortic insufficiency a relevant portion of the systolic aortic blood flow does not contribute to organ perfusion as it regurgitates into the left ventricle during diastole.

Whenever the basic assumptions needed to translate the measured Doppler frequency shift into CO are violated, the absolute displayed value may likely be incorrect. However, trend monitoring of CO should theoretically be possible as long as the basic conditions remain unaltered. Indeed, as described earlier, the usefulness of TED as a trend-monitoring device has been confirmed by validation studies in various patient populations. However, it is important to realize that changes which affect the basic assumptions of the measurement technique, such as aortic clamping or unaccounted changes in aortic diameter due to acute blood loss, may result in changes of the displayed CO value which do not necessarily reflect the true changes associated with these events. This emphasizes the need to realize the limitations of TED monitoring and to consider all available parameters and clinical signs in the assessment of the patient.

CONCLUSIONS

Transesophageal Doppler devices interrogate the descending aortic blood flow and calculate aortic blood flow velocity based on the measured Doppler frequency shift. The resulting velocity–time plot allows estimations of CO and other advanced hemodynamic parameters, making TED a technology for continuous hemodynamic trend monitoring. While its role in the treatment of critically ill patients and its potential to replace more invasive techniques has not sufficiently been established, numerous studies support its use for guidance of perioperative volume therapy. TED-guided volume optimization has been shown to reduce perioperative complications and length of hospital stay, suggesting that TED may be a valuable supplement to current standard hemodynamic monitoring.

Correct interpretation of TED-derived data does not only require a thorough understanding of cardiovascular

physiology, but also a profound knowledge of the technical basis of this method. Limitations may in part derive from inaccuracies in the determination of the blood flow velocity. However, especially the assumptions needed to translate descending aortic blood flow velocity to CO introduce a considerable potential for erroneous calculations. A simple ABCD scheme can be used to assess the validity of the displayed data:

Angle of insonation: known and constant?

Blood flow: laminar or turbulent?

Cross sectional area: unaccounted changes in aortic diameter?

Distribution of flow between supra-aortic vessels and descending aorta.

The user needs to be familiar with the inherent limitations to allow a rational and proper use of the technology.

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