

ORIGINAL PAPER

Color/pulsed Doppler measurement of renal blood flow in healthy children: a pilot study

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ABSTRACT

Introduction: The renal blood flow (RBF) is a major determinant of the kidney function, and its reliable measurement is potentially a relevant clinical tool for a functional assessment of the kidney. We used a Doppler blood flow measurement method, based on a simplified plug flow model, to assess RBF in healthy children.

Material and methods: The diameter and cross-sectional area of the left and right renal arteries, as well as the corresponding velocity-time integral (VTI) and RBF were measured in 41 healthy children, 21 girls and 20 boys, aged 4–17 years.

Results: The mean values for RBF and VTI were 1864 ml/min/1.73 m² (+/- 511) and 45.6 cm (+/-11.1), respectively. There was no significant difference of those parameters between sexes. Interestingly, when comparing sides, a better correlation was found between left and right RBF than for left and right VTI, with a Pearson correlation coefficient of $r = 0.903$ ($p < 0.001$) and $r = 0.72$, respectively ($p < 0.001$). There was no difference ($p = 0.13$) between RBF in our group and historical data based on renal clearances of para-amino hippurate. The mean RBF was also not higher than other data from the literature.

Conclusions: Applying a plug flow method to a laminar flow in an artery overestimates the blood flow. However, given the good left and right side correlation, the method may be clinically relevant in assessing the differential kidney function in children with obstructive uropathies, as well as in quantifying kidney hyperfiltration and stratification of the risk of progressive chronic kidney disease in children with solitary kidney.

KEY WORDS:

renal blood flow, Doppler ultrasound, children.

INTRODUCTION

The kidneys receive 20–25% of the cardiac output, which represents a blood flow of approximately 1 liter per minute in adults. Assuming the hematocrit constitutes roughly 40% of that volume, the average renal plasma flow is 600 ml per minute. The glomerular filtration fraction is about 20%, which gives 120 ml of filtrated plasma per minute, and represents the glomerular filtration rate. The renal blood flow (RBF) is obviously a major determinant of the kidney function, and its reliable measurement is po-

tentially a relevant clinical tool for a functional assessment of the kidney, similar for example to the differential renal function obtained in scintigraphy. However reliable methods of RBF measurement, like renal clearances of para-amino hippurate (PAH) or the radiocontrast material, diodrast, are invasive and impractical, especially in children. On the other hand, the assessment of the reliability of ultrasound blood flow estimation in adults was cautious [1]. Ultrasound blood flow estimation in adults is considered not entirely reliable primarily because calculating absolute volume flow is prone to operator-dependent variables and

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geometric errors (insonation angle, vessel's cross-sectional area, transversal vs. longitudinal methods). While velocity and relative changes are reliable, volumetric assessment struggles with several specific challenges [2]. Three difficulties arise when applying this blood flow measurement method to the renal artery. First, for a plug flow model, the insonation angle must be aligned to the renal artery. This is difficult to achieve in adults, but relatively easy in a substantial percentage of children. Secondly, the measurement of the renal artery diameter must be very precise, as every error is squared (πR^2), and multiplied by the velocity-time integral (VTI) and the heart rate (HR). Small variations of R result in massive changes in the calculated blood flow. The third difficulty is not associated with the technique per se, but with the important anatomical variants of the renal arterial vasculature [3]. A premise of this work was that it is possible to address those difficulties to an extent, and significantly minimize the measurement errors in a substantial percentage of children, with the aim to assess the possible clinical usefulness of those optimized measurements.

Taking advantage of the improvements in ultrasound imaging, as well as of the superior modality potential in children, we used a Doppler blood flow measurement method, based on a simplified plug flow model, to assess RBF in healthy children.

MATERIAL AND METHODS

The diameter and cross-sectional area (CSA) of the left and right renal arteries, as well as the corresponding VTI and RBF were measured in 41 healthy children, 21 girls and 20 boys, aged 4–17 years. The body surface area was calculated using Mosteller formula. Spectral velocity-time recordings were acquired with pulsed wave Doppler in triplex ultrasound. The velocity-time integral was obtained for one heart cycle. The cross-sectional area of the renal artery was carefully measured in B mode at the same site VTI was measured.

The exclusion criteria included the following: acute kidney injury, chronic kidney disease, congenital anomalies of kidneys and urinary tract. We also considered an accessory renal artery as well as an early branching if the bifurcation as an exclusion criterion in this work, made it impossible to measure the VTI and renal artery diameter in the common trunk. This was an important limitation due to the high occurrence of those variants, moreover those anatomic variants can be difficult to detect in ultrasound. A careful assessment of the number of renal arteries and their morphology is therefore mandatory.

There are several methods of measuring blood flow in ultrasound. The applied one involves the calculation of VTI in the renal artery and its CSA. The velocity-time curve shows the changing blood flow velocity in the renal artery as a function of time. The integral of this curve VTI, or in other terms the area under the curve calculated in cm

by the ultrasound machine, represents the distance covered by the blood in a time interval (1 heart cycle for instance). The cross-sectional area of the renal artery is next calculated by applying the measured artery diameter (D) to the π radius squared (πR^2) formula (with $R = D/2$). By multiplying VTI (the distance covered by the blood during one heart cycle) by CSA of the artery, a cylinder volume is obtained. This volume represents the blood flow during one heart cycle. The renal blood flow per minute is finally calculated by multiplying the obtained value by HR, according to the formula: $RBF = VTI \times (\pi R^2) \times HR$ (Figure 1).

We used instead the eFLOW (Aloka Co., Ltd., Tokyo, Japan) imaging mode in Aloka US machines, which offers a higher-definition blood flow imaging with improved spatial and temporal resolution thanks to widened bandwidth of the ultrasound transmitter/receiver.

In practice, to maximize the precision of the CSA measurement, we found it helpful to return to B-mode once VTI was obtained, then to zoom the frozen renal artery image, and select the best frame to carefully measure its diameter at the same point VTI was obtained.

Statistical tests were performed with IBM SPSS statistics v.26. Values are given as means $\pm$ SD. Normality was checked in Shapiro-Wilk test. Paired Student *t*-test and Pearson correlation coefficient were used when comparing left and right kidney parameters. Difference between RBF in our group and historical RBF values were tested with Welch *t*-test, with respect to the different population and variance. The significance level was assumed as *p*-value < 0.05.

RESULTS

The mean values for RBF and VTI were 1864 ml/min/1.73 m² (± 511) and 45.6 cm (± 11.1), respectively (Figure 2). There was no significant difference of those parameters between sexes. Interestingly, when comparing sides, a better correlation was found between left and right RBF than for left and right VTI, with a Pearson correlation coefficient of $r = 0.903$ ($p < 0.001$) and $r = 0.72$ ($p < 0.001$), respectively (Figure 3). There was no difference ($p = 0.13$) between RBF in our group and historical data based on renal clearances of PAH by Krüge Brodwall [4] (Figure 4).

DISCUSSION

Blood flow velocity in the renal artery, as in most arteries, is not homogenous throughout the vessel cross section. It is usually laminar, with faster flow in the center and slower toward the vessel walls. The velocity profile is accordingly parabolic in shape (Figure 5 A), as opposed to the plug flow model (Figure 5 B). The latter assumes a flattened, near linear velocity profile and is observed in large vessels like the aorta, or at the entrance of a vessel. Velocity profiles are complex and difficult to model

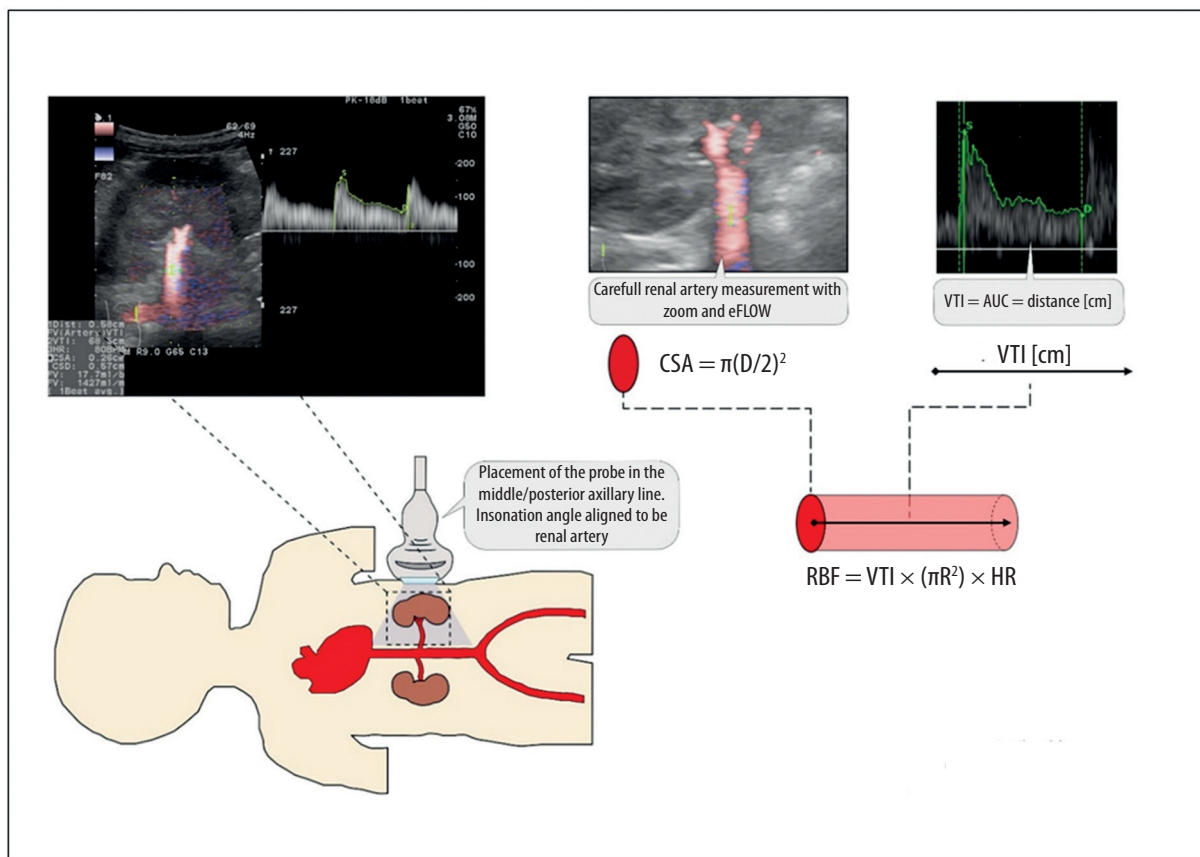


FIGURE 1. Plug flow Doppler blood flow measurement applied to the renal artery

AUC – area under the curve, CSA – cross-sectional area, D – diameter, HR – heart rate, RBF – renal blood flow, VTI – velocity time integral

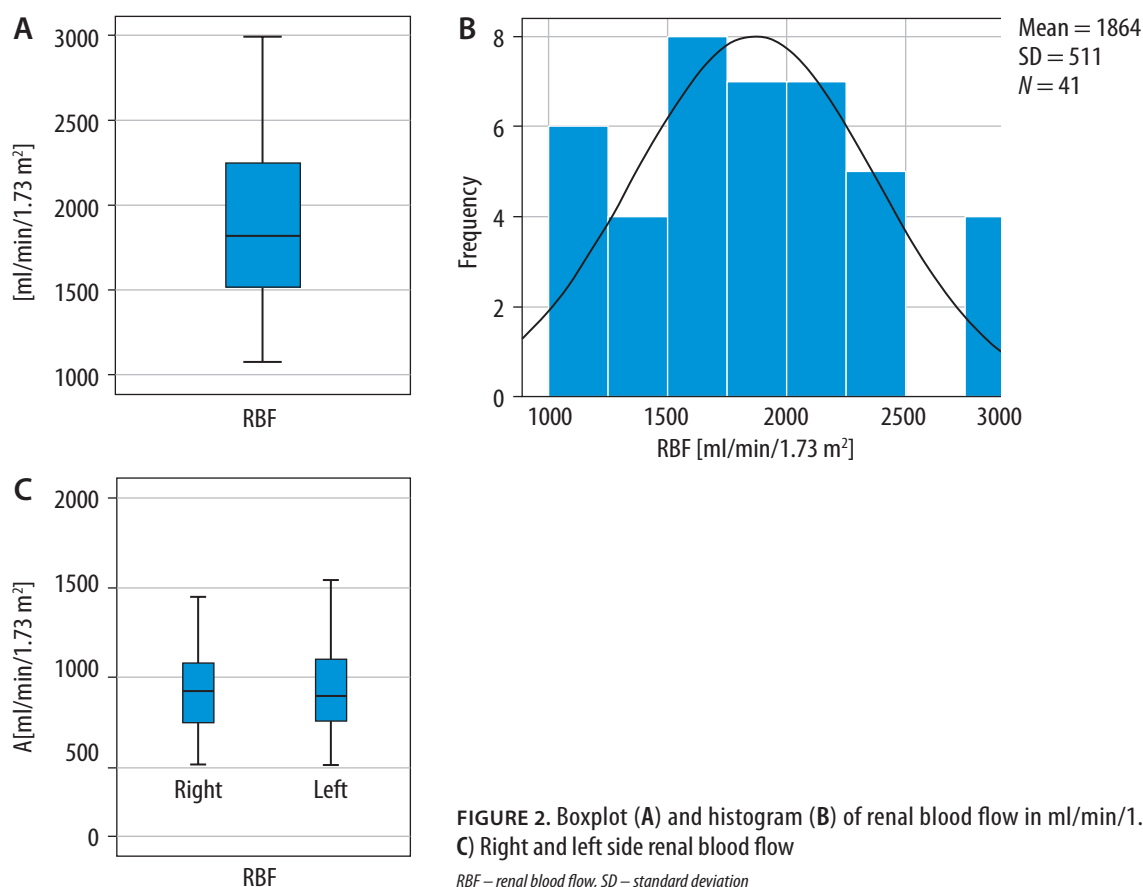


FIGURE 2. Boxplot (A) and histogram (B) of renal blood flow in ml/min/1.73 m². C) Right and left side renal blood flow

RBF – renal blood flow. *SD* – standard deviation

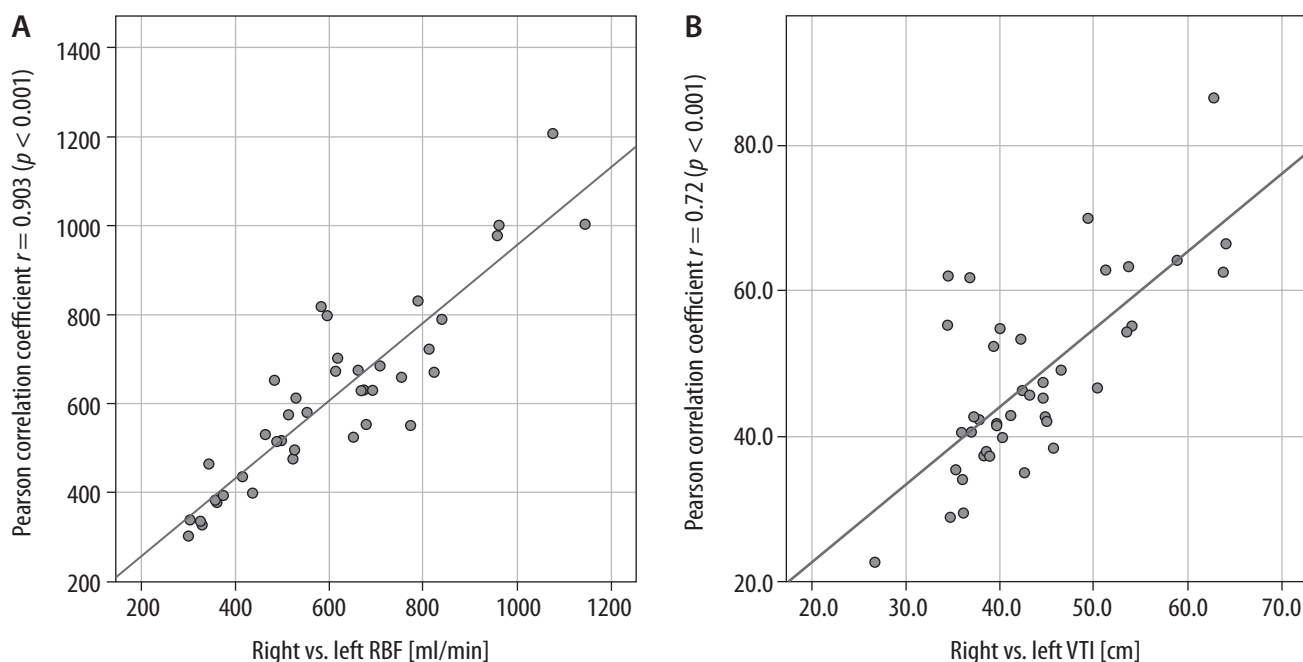


FIGURE 3. Pearson correlation coefficient and scatterplot right vs. left renal blood flow and velocity time integral

RBF – renal blood flow, VTI – velocity time integral

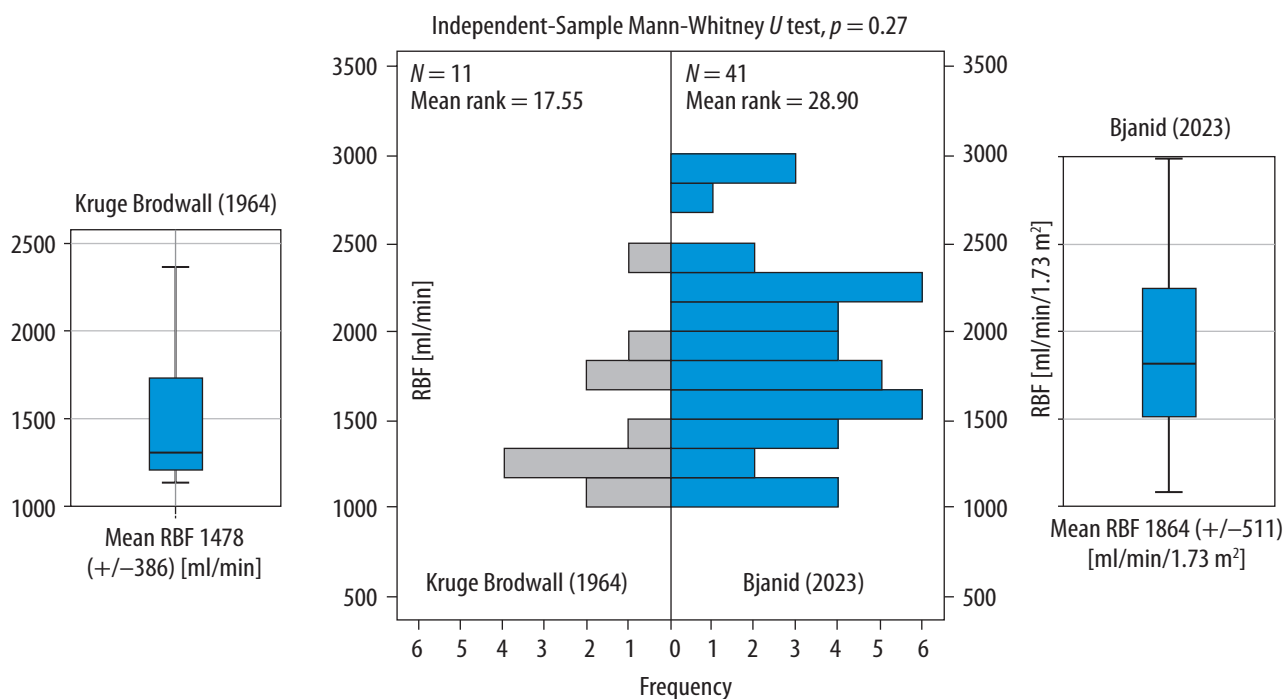


FIGURE 4. Renal blood flow in our group vs historical data based on renal clearances of para-amino hippurate

RBF – renal blood flow

as they are affected by numerous factors like the distance from bifurcation, vessel curving, bending and moreover, pulsatile by definition, they change dynamically during systole and diastole as depicted in Figure 5 C. A more detailed description of velocity profiles in the renal arteries can be found in the paper by Yamamoto *et al.* [5].

As mentioned, there are different methods of measuring blood flow in Doppler ultrasound depending on

the laminar or plug flow pattern in the artery of interest. For laminar flow, the ultrasound beam should cross the artery at an angle, and the Doppler sample volume must encompass the entire vessel cross section (Figure 5 D). The average velocity is then calculated and multiplied by the cross section. This requires a precise placement of the sample volume at the vessel walls, which was virtually impossible to achieve in our group, due to the small diameter

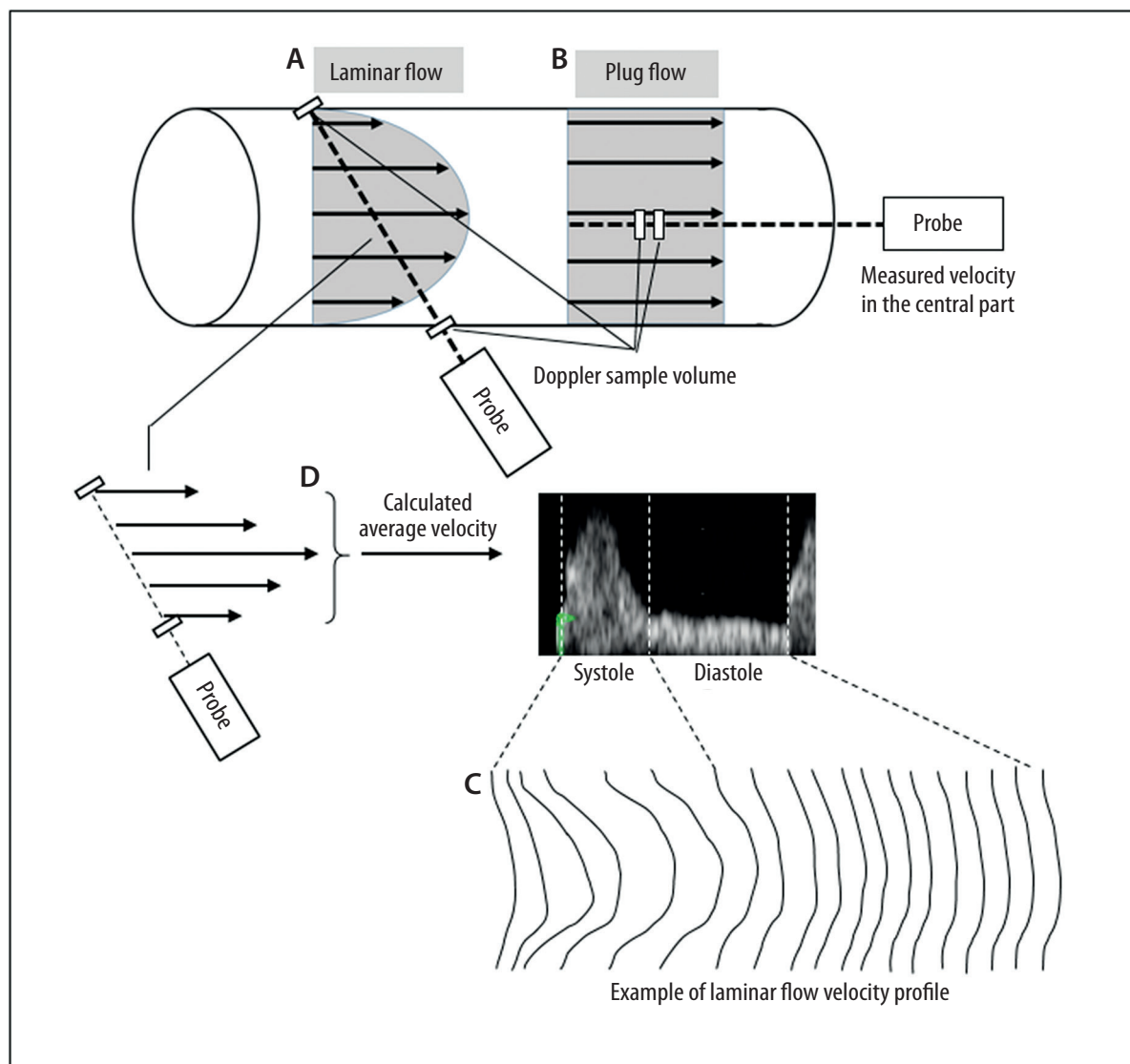


FIGURE 5. A) Laminar flow. B) Plug flow. C) Velocity profile in renal artery. D) Doppler blood flow measurement in laminar flow

of the renal artery in children, the respiratory movements, and the requirement of a very precise CSA measurement, which we performed on a frozen and zoomed image. We used instead a simplified, plug flow approach, as described in the methodology. A more detailed review of the technical aspects of Doppler blood flow measurement can be found in vascular ultrasound literature, for instance the comprehensive review by Gill [6].

The issue of aligning the insonation angle to the renal artery is relatively easily addressed in children by positioning the transducer in the posterior axillary line, along the kidney long axis, and tilting it towards the aorta as shown in Figure 1. This enables a good visualization of the renal artery in a majority of children. A helpful way of aligning the renal artery to the insonation angle is to ask the child to take a deep breath and exhale slowly, until the angle reaches 0° , the child to hold breath (Figure 6). If properly explained and even shown on the screen, most children in our experience were able to cooperate and, with some training, it was possible to obtain an adequate

visualization in a majority of cases. Further fine tuning can be achieved by rocking the transducer, that is applying a motion in the long axis of the probe along a fixed point on the body while changing the angle of insonation away from 90° , as defined by the American Institute of Ultrasound in Medicine [3].

As stated before, one of the greatest difficulties, if not the major drawback to the method, is the necessity of a very precise measurement of the renal artery diameter (D). Small changes in the measured diameter result in very important variations in the calculated blood flow. For an average VTI of 42 cm, and an average HR of 80 bpm, a +1, +0.5, -0.5 and -1 mm change in D, results in a change in measured RBF of +44%, +20%, -18% and -35%, respectively (Figure 7). To stay below a margin of error of 20%, the renal artery diameter must be measured with a resolution better than 0.5 mm. This is not achievable with conventional color or power Doppler because of the blooming or "bleeding" artifact that makes the color "spill out" of the artery, and exaggerates its diameter.

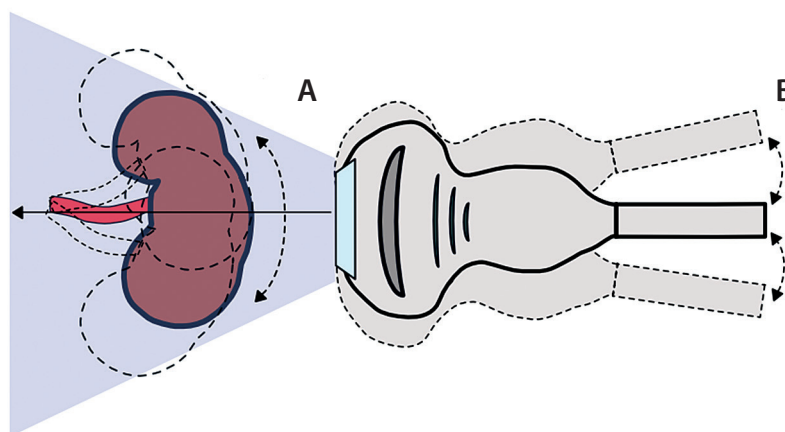


FIGURE 6. Inhalation and expiration (A) and transducer rocking (B) help in achieving a 0° angle between the ultrasound beam and the renal artery

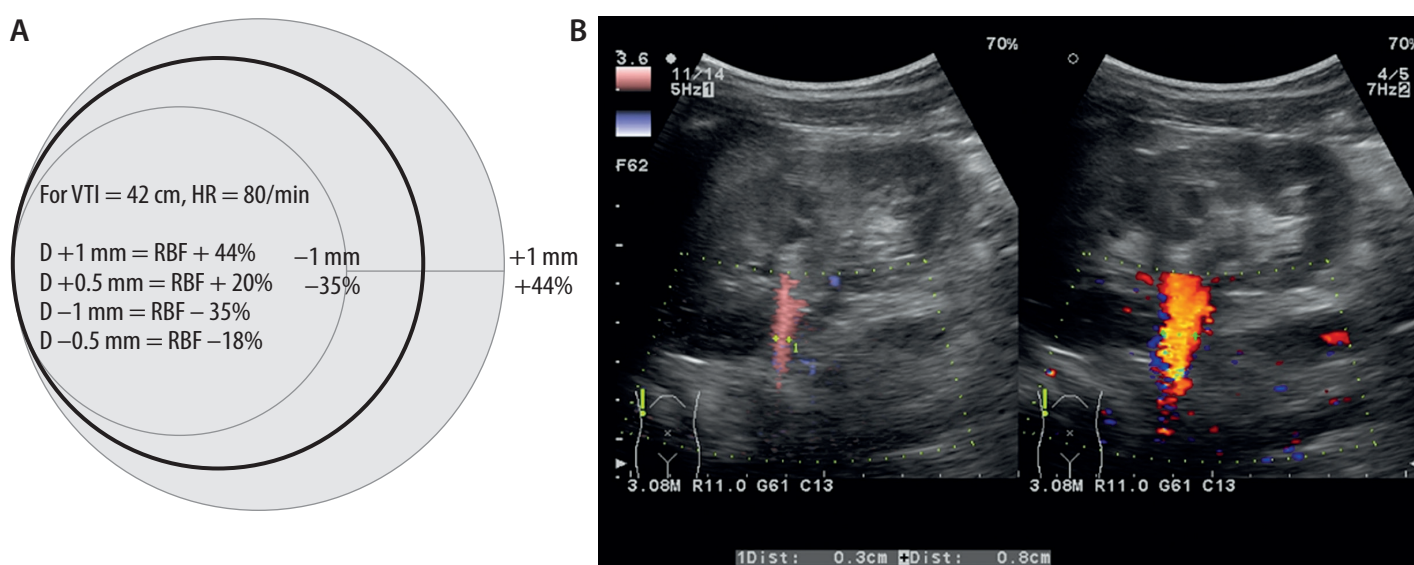


FIGURE 7. A) Small errors in the renal artery diameter (D) measurements result in important fluctuations in renal blood flow. B) Blooming or “bleeding” artifact in conventional color Doppler (right) vs. eFLOW imaging mode (left)

RBF – renal blood flow

A final difficulty consists in the relatively high occurrence of renal artery anatomic variants. In a study on 314 kidney donors who underwent 64-channel multi-detector computed tomography angiograms, Song *et al.* [7] found an accessory renal artery in 25% and 19.4% of cases on the right and left side, respectively. Furthermore, an early branching of the right and left renal artery was found in 13.7% and 10.5%, respectively of the patients.

In summary, we measured the maximal velocity at the central part of the renal artery, and applied it to the entirety of the vessel cross section, which exaggerates the average cross-sectional velocity and subsequently the calculated blood flow. This is consistent with the lack of difference between our RBF measurements and those based on renal clearances of PAH as reported by Williams *et al.* [8] and Takano *et al.* [9]. However, the cohort evaluated by these authors had a low number of participants. The cited authors reviewed and analyzed data on absolute blood flow rates to various organs. They pointed that flow

measurement using Doppler ultrasound may be a reliable noninvasive method for quantifying RBF when measured values exceed 280 ml/min per 1.48 m².

However, even if a precise volumetric Doppler RBF measurement seems out of reach, we believe the described method is still useful in a number of frequent clinical situations in pediatric nephrology. For instance, the good right and left RBF correlation offers an opportunity to assess renal function asymmetry in the context of an obstructive uropathy. The management of those hydronephrosis conditions is based upon such a functional assessment in renal scintigraphy. A Doppler RBF measurement could help identify patients at risk, in whom scintigraphy should be performed, as well as provide a relatively simple, non-invasive, follow-up modality. Another application of the described method is the quantification of hyperfiltration in solitary kidney or other conditions with reduced kidney mass, and the risk stratification of progressive chronic kidney disease in those

children. We are currently assessing the method in both groups of children with a solitary kidney, and asymmetry of differential kidney function in scintigraphy. There are many other clinical settings in which RBF measurement may be a valuable input, like in acute kidney injury, in the transplanted kidney or in thrombotic microangiopathies and more data are needed to assess this modality.

As already mentioned there are limitations to the described method. Beside the inherent overestimation of blood flow, it can be technically challenging and requires experience in kidney Doppler ultrasound. We have been routinely performing them for more than ten years, mainly to rule out renal stenosis in hypertensive patients. In our experience the major sources of errors are the precision of the CSA measurement, the alignment of the insonation angle to the long axis of the artery, the placement of the Doppler sample volume in the center of the vessel, and finally the frequent anatomic variants, which must be carefully ruled out.

The recent advances in ultrasound are impressive. The excellent spatial resolution offered by modern ultrasound devices, like the e-flow modality, opens new opportunities for blood flow measurement. Other advancements like the possibility of a simultaneous dual-point Doppler velocity measurement, make possible better quantification of the described overestimation error, and further refinement of the measurement, based on velocity profile knowledge. Novel devices like hand-free patches for continuous monitoring of the blood flow in the carotid artery [10] or microvascular systems for intraoperative blood flow evaluation are being introduced. The field of point-of-care sonography is evolving so dynamically, that things considered impossible, like an assessment of the kidney function in ultrasound are, quite literally, at the tips at the clinician's fingertips.

CONCLUSIONS

Applying a plug flow method to a laminar flow in an artery overestimates the blood flow. However, given the good left and right side correlation, the method may be clinically relevant in assessing the differential kidney function in children with obstructive uropathies, as well as in quantifying kidney hyperfiltration and stratification of the risk of progressive chronic kidney disease in children with solitary kidney.

DISCLOSURES

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2. Assistance with the article: None.
3. Financial support and sponsorship: This study was supported by research grant number KNW-1-073/N/8/K from the Medical University of Silesia.
4. Conflicts of interest: None.

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