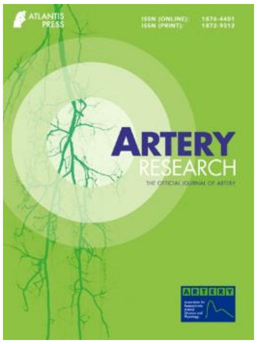

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P6.11: EVALUATION OF A NOVEL AND EXISTING TECHNIQUES FOR THE ESTIMATION OF PULSE TRANSIT TIME

O.V. Vardoulis, T.G.P. Papaioannou, N.S. Stergiopulos

system and (2) a non-distensible zone, disturbing the buffer function of the aorta. As the many interfering factors and adaptive physiologic mechanisms present in vivo prohibit the study of the isolated impact of these individual factors, an advanced computer model was developed.
Material and methods: The geometry and flow boundary conditions are obtained from MRI data of a healthy subject (Figure 1). A segment with varying length and stiffness was included distal to the left subclavian artery (red zone in Figure 1). Recurrent coarctation was studied by altering the diameter (coarctation index of 0.5 for severe and 0.65 for mild coarctation).
Results: Figure 2 depicts the effect of a local non-distensibility on the pressure evolution proximal and distal to the rigid zone. Data shown represent the presence of a stent (length 5cm, 100 x stiffer than reference material) or scar tissue (length 5 mm, 5x stiffer). Although the overall impact is very limited, the presence of a stent increased the proximal systolic pressure with 4.5 mmHg compared to the pressure in a healthy subject.
Conclusion: The model allows to study the isolated effect of local non-distensibility and narrowing which is impossible to obtain in vivo.

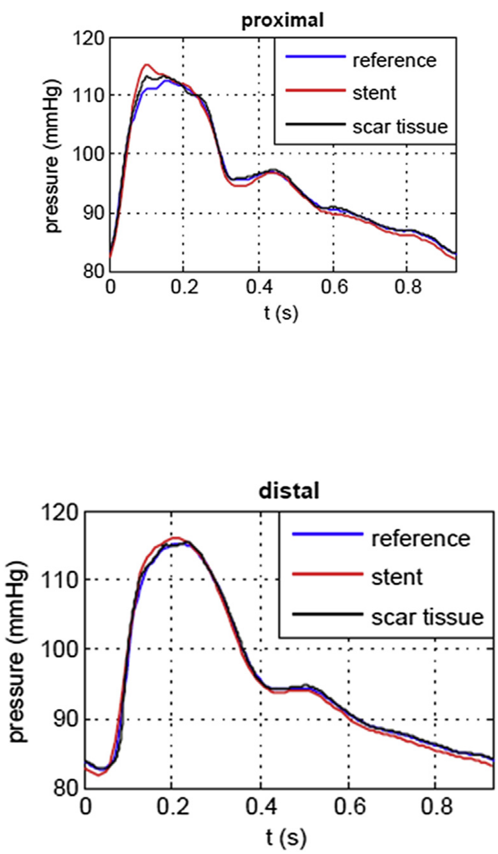

Figure 2 Proximal and distal pressure.

P6.09
MULTIPLE REFLECTIONS, NOT A SINGLE DISTAL AORTIC REFLECTION DETERMINE PRESSURE WAVE SHAPE

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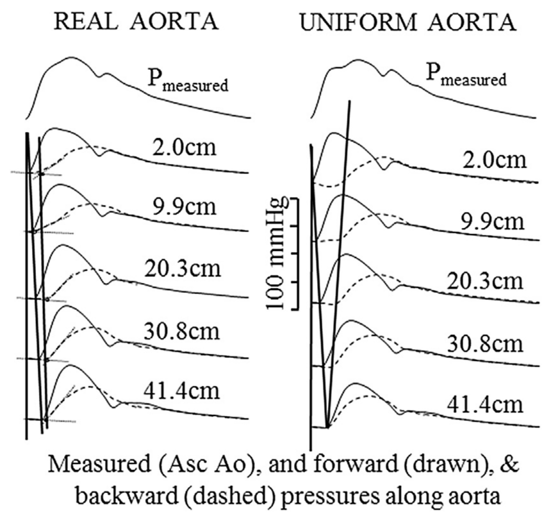
Arterial pressure and flow waves travel and are reflected. Waveform analysis and wave separation gave insight into these phenomena and parameters thus obtained are indicators of cardiovascular events. However, the interpretation of forward and reflected waves is still not generally agreed upon. We used an anatomically accurate (data from Hickson, 2010) model of the entire systemic arterial tree and also set all aortic diameters equal at mean aortic diameter ("uniform aorta"), leaving other arteries unchanged, and calculated forward and backward waves in the standard way (Murgo, 1981). In the anatomically accurate model, timing of the feet of backward and forward waves is location independent, as also recently reported by Tyberg, 2013. In the uniform aorta the delay between forward and backward waves is smallest in the distal aorta and largest in the ascending aorta. In both models pressure amplification over the aorta is ~1.35. Changes in microcirculatory resistance have little effect on wave shapes. We conclude that multiple local reflections in the aorta importantly contribute to pressure (and flow) wave shape. Thus pressure wave shapes depend on arterial geometry: aortic diameters and side branches. Distal aortic (bifurcation) and peripheral reflections are not the major contributors to overall reflection and wave shape. We suggest that studies of aortic dimensions and effect of side branches are needed to better understand aortic pressure wave shapes and wave travel.



Hickson et al., JACC Cardiovas Imaging 2010;2:1247.
Murgo JP et al., Circulation 1981; 63, 122.
Tyberg JV et al. J of Physiol 2013;591.5 p 1171.

P6.10
SENSITIVITY OF WAVE SEPARATION IN THE ARTERIES TO ERRORS IN ESTIMATING WAVE SPEED

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Objectives: Examine the effect of erroneous estimation of wave speed (C) on the magnitude and time of the separated pressure (P) and velocity (U) in arteries.
Methods: Pressure and flow were measured in the aorta of 11 dogs and C has been determined using the PU-loop technique. The waves were separated into the forward and backward directions using wave intensity analysis (WIA), with C varying from C-99C% to C+99C%. The following parameters were studied: **a)** Peak of forward (P+, U+) and backward (P-, U-) pressure and velocity waveforms, **b)** The onset and peak times of P+, U+ , P-, and U- all with respect to ventricular ejection time.
Results: Incorrect values of C resulted in an inaccurate estimation of the $P_{\pm}$ and $U_{\pm}$. An error of (+,-)50% in C results in an amplitude error of 7,7% in P_{+}, 6, 8% in P-, 20, 60% in U+ and 30, 116% in U-. Also, an error of (+,-) 50% in C results in an error in peak time of 7, 11% for P+, 15, 5% for P-, 7, 10% for U+ and 2, 20% for U-. Incorrect determination of C did not affect the onset of the forward waves while it resulted in error of 47,47% for P- and 38,38% for U- (Figure 1,2).
Conclusions: The separation of P and U waveforms using WIA is sensitive to changes of C, whose correct estimation is important for the accurate determination of the magnitude and peak time of the forward and backward waveforms.

P6.11
EVALUATION OF A NOVEL AND EXISTING TECHNIQUES FOR THE ESTIMATION OF PULSE TRANSIT TIME

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The method used for pulse transit time (PTT) estimation, affects critically the accuracy of pulse wave velocity (PWV) measurements. The existing methods for PTT estimation yield often substantially different PWV values. Since there is no analytical way to determine PTT in vivo, these methods cannot be validated except by using in silico or in vitro models of known PWV and PTT. We aimed to validate and compare the most commonly used "foot-to-foot" methods: "diastole-minimum", "tangential", "maximum 1st derivative" and "maximum 2nd derivative". Also, we propose a new "diastole-patching" algorithm aiming to increase the accuracy and precision in PWV measurement. Methods: We simulated 2000 cases under a range of different hemodynamic conditions using a validated, distributed 1-D arterial model. The new algorithm "matches" a specific region of the pressure-wave foot between the proximal (i.e. carotid) and distal (i.e. femoral) waveforms. Intraclass correlation coefficient (ICC), mean difference (bias) and standard deviation of differences (SDD) were used to assess accuracy and precision. Results: The " diastole-minimum" and the " diastole-patching" methods showed an excellent agreement compared to the "real" PWV values of the model, as indicated by high values of ICC(>0.86).
The "diastole patching" method resulted in low bias (0.26m/s). In contrast, PWV estimated by 1st or 2nd derivatives and the "tangential" method presented a low to moderate agreement and poor accuracy (ICC<0.79, bias>0.9 m/s). The "diastole-patching" method yielded PWV measurements with the highest agreement, accuracy, precision and the lowest variability and its validity remains to be further examined in vivo.

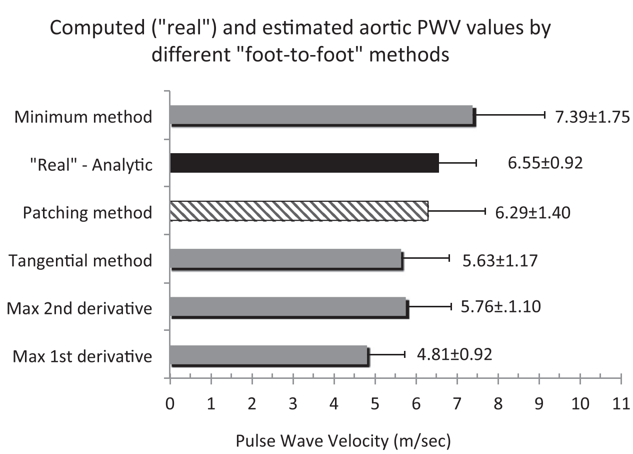

Figure 1 Mean and standard deviation for the aortic PWV estimations of the 5 validated algorithms. Bar in black represents the "real" PWV of the model.

P6.12

WAVE INTENSITY ANALYSIS OF REFLECTIONS IN THE BRACHIAL ARTERY FOLLOWING CUFF OCCLUSION AND HAND WARMING

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Background: Wave intensity analysis (WIA) is a mathematical tool used to study wave reflections in the arteries. Reflections are believed to contribute to BP augmentation and are also independent predictors of cardiovascular risk. Until now, the use of this technique has been largely confined to the aorta and carotid arteries.
Methods: 8 healthy subjects (age 30 ± 7.1) underwent wrist occlusion using a cuff inflated to >50mmHg suprasystolic pressure for 5min and hand warming at 55°C for 12min. Brachial artery diameter and blood flow velocity were measured using wall tracking and doppler ultrasound with an ALOKA SSD-5550 equipped with a 7.5 MHz probe. Wave intensity was calculated and reflections were quantified as the energy of the reflected wave/energy incident wave (WRI, %). Central aortic pressure following hand warming was also estimated using applanation tonometry (Sphygmocor) in separate studies.
Results: Cuff inflation resulted in a significant increment in WRI from 12.4 ± 4.15% to 26.8 ± 8.34% (p=0.001) whereas a marked reduction from 16.3 ± 6.60% to 4.09 ± 1.62% followed hand warming (p=0.0017). Cuff release was immediately associated with a significant attenuation in WRI (p=0.01). Hand-warming had no significant effect on the contralateral brachial or aortic SBP or DBP compared to baseline (p=0.3, 0.08 respectively).
Conclusion: Radial artery occlusion and hand warming respectively led to an augmentation and reduction in the reflected wave in the brachial artery. Hand warming was not associated with a significant change in peripheral or central BP.

P6.13

ADULT GUIDE-LINES ARE NOT APPLICABLE TO MEASURE PWV PATH LENGTH IN PAEDIATRICS

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Aortic pulse wave velocity (PWV) is a sensitive marker of arterial stiffness in children. In our previous study we have presented reference tables for PWV normal values in children. A recent consensus document provides arguments for the use of 80% of the direct carotid femoral distance as the most accurate distance estimate in adults. In the present work we aimed to assess if a transposition of the adult PWV measurement method is valid in childhood. Data of children participating to our previous work establishing age and height specific PWV normal values were re-evaluated. A total of 1008 healthy children (mean age:15.2 years, 495 males) were included in the study. We have recalculated PWV values using the subtractive method path length (L(SM)) and 80% of direct path length (L(0.8)). We have constructed Bland-Altman (BA) plots to assess the difference between PWV(SM) and PWV(0.8), and the distances L(SM) and L(0.8) in different age groups. The concordance between PWV(SM) and PWV(0.8) is excellent in children below 14 years (BA, ΔPWV mean:0.19 m/s, SD:0.40).However, in children >14 years, the difference increases (BA, ΔPWV mean:0.57 m/s, SD:0.36), and there is a proportional error between PWV(SM) and PWV(0.8) (BA, r:0.18; p<0.001), and in parallel there is also a proportional error between L(SM) and L(0.8) (BA, r:-0.24;p<0.001). The path length measurement suggested for adults may not be transponible to children throughout all age groups without reservation. Thus we propose to keep the current tables and values, unless the validity of a particular measurement is proved.
(Grant: OTKA100909)

P6.14

THE EFFECT OF TEMPORAL RESOLUTION ON MR ASSESSMENT OF PULSE WAVE VELOCITY

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Objectives: PWV can be measured by velocity-encoded phase-contrast magnetic resonance imaging (PC-MR) in a single location. One method utilises the change in flow (ΔQ) divided by the change in area (ΔA) at the beginning of systole, when it is assumed that only forward running waves are present. However, the duration of a reflection free period is short (~30ms) and therefore a high sampling frequency is required to interrogate this period. Most PC-MR is performed with a low TR of approximately 30-40ms. In this study, we compared PWV calculated using high TR (10ms) and simulated low TR (30ms).
Methods: High TR (10ms) PC-MR was performed in 20 volunteers in the ascending aorta. TR reduction to 30ms was simulated by filtering the flow and area waveforms using a zero-phase, low-pass, high-order Butterworth filter with normalized cut-off frequency of 0.33 in Matlab. PWV was calculated from the gradient of the flow-area line at the onset of ejection, corresponding to the first 3 points of the foot of the area curve.
Results: There was a significant difference (p<0.0004) between PWV calculated using high TR mean 3.89m/s (SD 1.31) compared with simulated low TR, mean 7.30m/s (SD 3.64), Figure A. The mean bias between methods was 3.4m/s with wide limits of agreement (Figure B).
Conclusion: PWV calculated using a single location method is significantly inflated when data is acquired at low TR, as simulated using low-pass filtering. This suggests that conventional PC-MR may produce erroneous results in clinical studies.