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Non-invasive estimation of the pulmonary capillary pressure with echo-Doppler in patients without left ventricular dysfunction

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Background and objectives: Different studies have validated the non invasive estimation of the pulmonary capillary pressure (PCP) with Doppler Tissue Imaging (DTI) and colour Doppler M mode. In patients (pat) with left ventricular (LV) dysfunction, however, the application of these methods has not been well studied in patients with preserved LV function nor it has been compared side by side. The aim of this study was to validate these methods in patients with normal LV ejection fraction (EF).

Methods: An echo-Doppler and a right cardiac catheterization were performed on the same day in 33 patients with liver cirrhosis and no previous cardiovascular disease. LV diameters, LVEF and early peak LV filling flow were evaluated with M-mode/2D echo and with pulsed Doppler (E) and colour Doppler M-mode (Vp); the peak early diastolic velocity of the lateral mitral annulus was also determined with pulsed Doppler (E′). PCP was estimated according to 3 previously validated formulas in pat with LV dysfunction: PCP1 = 5.27 x ESVp + 4.66; PCP2 = 1.9 x 1.24 x E/E′; PCP3 = 1.65 + 1.47 x E/E′. LV cardiac output (CO) and PCP were measured with a Swan-Ganz catheter. The regression (regression analysis) and concordance (Bland Altman analysis) between Doppler-estimated PCP and the one obtained by catheterization was analyzed.

Results: Mean LV EF calculated by echo was 65±5% and mean LV CO and PCP obtained by catheterization were 8.6±2.3 l/min and 10.0±4.4 mmHg, respectively. The table shows PCP mean values estimated with the 3 formulas and the results of the regression and concordance analysis referred to PCP obtained by Swan Ganz catheter.

Table 1

<table>
<thead>
<tr>
<th>Mean±SD (mmHg)</th>
<th>Pearson R coefficient</th>
<th>p</th>
<th>Mean Error (mmHg)</th>
<th>Absolute mean error (mmHg)</th>
<th>Concordance limits (mmHg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PCP1</td>
<td>10.7±1.5</td>
<td>0.14</td>
<td>0.4</td>
<td>0.6±1.6</td>
<td>3.5±12.9</td>
</tr>
<tr>
<td>PCP2</td>
<td>7.2±1.8</td>
<td>0.61</td>
<td>-0.01</td>
<td>2.8±3.9</td>
<td>3.0±3.5</td>
</tr>
<tr>
<td>PCP3</td>
<td>7.8±2.2</td>
<td>0.61</td>
<td>-0.01</td>
<td>2.1±3.7</td>
<td>3.3±2.7</td>
</tr>
</tbody>
</table>

Conclusions: Our study demonstrates that the non-invasive PCP estimation in patients with normal LV function: (a) is feasible and precise when compared with Doppler echocardiography and (b) is more accurate using echocardiographic methods based on DTI than those derived from colour M-mode.

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Fluid dynamics study of three-dimensional left ventricular blood flow patterns

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A mathematical model was developed describing the normal flow pattern in the left ventricle (LV). The 3D Navier-Stokes equations ruling intraventricular fluid mechanics were solved numerically in a LV half prolate spheroid geometry with moving LV walls. The computed results were compared with actual data obtained from 2D Doppler color flow mapping echo performed on healthy human subjects.

Results: The mathematical model of 3D flow pattern within the LV demonstrated that LV entry inflow is not directed axially to the apex, but it is mainly redirected toward the lateral wall. It develops a vortex that fills the LV cavity, blood flows toward the apex on one side and round the anterior mitral leaflet on the other side. This circulation corresponds to a slice of a 3D vortex structure shaped as an incomplete vortex ring (figure 1). These data account for the swirling nature of the LV flow detectable at 2D Doppler color flow velocity mapping echo. As a consequence of the asymmetry of the mitral inflow the apparent slowdown in the color M-mode Doppler tracing is not due to a progressive decay of the inflow jet velocity, but to the fact that higher velocity regions progressively move away from the M-mode plane. From the physiological point of view, color M-mode Doppler tracing does not represent "blood propagation velocity" from LV base to apex.

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An automated method to measure ultrasound signal quality

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Beat-to-beat variability, artifact and noise can all reduce signal quality in ultrasonic recordings, and therefore reproducibility. We tested 3 different methods of assessing signal quality automatically, using data recorded to measure net wave intensity (WI).

Methods: Velocity and local arterial pressure (calibrated from M-mode diameter) were recorded simultaneously from the right common carotid artery, using an Aloka SSD-5500 ultrasound scanner. WI was calculated as the product of derivatives of pressure and velocity, and a pressure/velocity loop was displayed. Using the water hammer equation, local wave speed was calculated for linear regions during early and late systole (coinciding with unidirectional wave travel), and also averaged throughout systole by matrix division. Ratios of the different wave speeds, plus a measure of signal noise for pressure, velocity and wave intensity (differences between each original vector and a low-pass filtered version) were assessed to measure noise. These indices were compared in 60 studies identified as good-quality by an experienced observer, and 35 other studies.

Results: Mean values and standard deviations of the 5 quality indices were calculated. Signals with good quality showed little variation (e.g. velocity and WI traces, see figure, upper panels) whereas routine recordings with suboptimal quality showed skewed and greater variability in these indices (figure, lower panels; the vertical lines represent mean±2sd for good-quality data)(all comparisons, p<0.05).

Conclusion: The model exhibits linearity regarding the dynamic properties in comparison to the reference. This indicates that the model is suitable in validation studies of TDI.

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Simulating animal targeted tissue acoustic behaviour to enable contrast agent development and study

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Several materials have been recently used to manufacture tissue mimicking phantoms for ultrasound (US) studies with flowing contrast agents.

Aim: To evaluate the acoustic behaviour of a new cellulose-based hydrogel, developed to simulate animal targeted tissue, through the comparison with an animal tissue.

Methods: A commercially available echograph equipped with a linear transducer and linked to a prototype for radiofrequency (RF) analysis was employed to insinuity both hydrogel and animal tissue samples with 10-cycles US pulses at different frequencies (2, 2.5 and 3.3 MHz). Echograph electrical power was varied for each frequency to obtain different mechanical index (MI) values, estimated assuming an attenuation of 0.3 dB/cm/MHz. For both samples, a sequence of 10 frames of RF raw data was acquired for each instrument setting and analyzed off-line. A rectangular region of interest (ROI) of approximately 5 mm² was selected and Mean Fast Fourier Transform curve was calculated within the selected ROI for each acquired frame. Then subharmonic (1/2H), fundamental (1H) and second harmonic (2H) component values were extracted from the obtained curves, averaged over the corresponding frame sequence and plotted versus MI.

Results: At every tested frequency, hydrogel backscattered intensity curves show the same trends of the corresponding animal ones (as a first step the liver). In particular, at 2 MHz, fundamental and subharmonic hydrogel curves show the same corresponding values of the liver (Fig. 1).

Conclusion: Cellulose-based hydrogel is a suitable material to manufacture a tissue mimicking phantom for US studies, since it well reproduced the echogenic properties of the targeted tissues with the investigated frequencies, chosen in the range of typical clinical studies.
S182 Abstracts

1126 Semi-automated epicardial border tracking
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Introduction: Endocardial semi-automated tracking has been developed recently. As assessment of left ventricular function is dependent on wall thickening rather than endocardial excursion this has limitations. We assessed a software application (Quamars) for semi-automated epicardial tracking.

Methods: Four dobutamine contrast stress echocardiograms of apical 4,3,2-chamber and short axis views at rest, intermediate and peak stress were obtained. Epicardial borders were hand-drawn by two experienced stress echocardiographers and also using Quamars in 288 frames for rest, intermediate and peak stress. Frames by frame a total of 3150 regions were assessed. 100 corresponding points were found for each pair of contours using symmetric nearest neighbours and arc length linear interpolation. The mean (dmean) and maximum (dmax) distances between correspondents were used as a measure of agreement between contours. The parameters of the lognormal (In) distribution μ and σ were calculated. The mean and standard deviation of the normal distribution of ln(dmean) and ln(dmax) were calculated.

Results: The distribution of dmean and max are lognormal. μ and σ are shown in the table below. Distances of dmean and dmax were very similar, as can be seen in the table. A Kolmogorov-Smirnov test showed the distribution of Quamars and one expert to be the same as between both experts in all stages of stress.

<table>
<thead>
<tr>
<th>Stage</th>
<th>dmean (mm)</th>
<th>dmax (mm)</th>
<th>μ</th>
<th>σ</th>
<th>μ</th>
<th>σ</th>
</tr>
</thead>
<tbody>
<tr>
<td>Obs 1 v Quamars</td>
<td>Rest</td>
<td>0.469</td>
<td>0.443</td>
<td>1.458</td>
<td>0.382</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Int</td>
<td>0.537</td>
<td>0.451</td>
<td>1.457</td>
<td>0.401</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Peak</td>
<td>0.628</td>
<td>0.456</td>
<td>1.562</td>
<td>0.434</td>
<td></td>
</tr>
<tr>
<td>Obs 2 v Quamars</td>
<td>Rest</td>
<td>0.653</td>
<td>0.420</td>
<td>1.627</td>
<td>0.386</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Int</td>
<td>0.702</td>
<td>0.420</td>
<td>1.805</td>
<td>0.399</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Peak</td>
<td>0.786</td>
<td>0.456</td>
<td>1.711</td>
<td>0.438</td>
<td></td>
</tr>
<tr>
<td>Obs 1 v Obs 2</td>
<td>Rest</td>
<td>0.477</td>
<td>0.487</td>
<td>1.464</td>
<td>0.457</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Int</td>
<td>0.611</td>
<td>0.415</td>
<td>1.559</td>
<td>0.395</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Peak</td>
<td>0.548</td>
<td>0.441</td>
<td>1.528</td>
<td>0.424</td>
<td></td>
</tr>
</tbody>
</table>

Conclusion: Computerised epicardial tracking is feasible and has the potential to allow quantification of left ventricular function.

1127 Estimating volumetric flow by exploiting the dependency of the Doppler spectrum on the position and insonation direction of an intra-vascular Doppler wire. An in-vitro study

Introduction: Commercial intra-vascular Doppler wire (IVDW) systems only estimate flow velocities and are highly dependent on wire position. Measurement of volumetric flow velocities and are highly dependent on wire position. Measurement of volumetric flow (Q) can be assessed through signal processing. The accuracy of these algorithms, however, highly depends on (knowledge of) the position of the DW, and on the potential disturbing effect of the DW on the flow velocity profile (VP) itself. The aim of this numerical study was to assess the effect of a perfectly centered DW and un-centered DW on VP and PDS.

Methods and Materials: DvC and DWoc were simulated in arteries with diameters of 3 (AD3) and 4 mm (AD4), respectively. In all cases, a constant Q of 215 ml/min was imposed on the walls of the vessel and the DW (no-slip condition). Cross-sections were defined every 0.13 mm (range gate step) perpendicular to and starting from the tip of the DW until 2 cm distal from the tip. Simulations were performed with the CFD (computational fluid dynamics) solver Fluent 6.1. Velocity data were extracted and subsequently used for further post-processing. With the opening angle of the ultrasound beam known (13°), the area sampled by the DW was determined for each range gate and velocity data were converted into PDS.

Results: For DWc, VP immediately distal to the tip has a "doughnut" shape, with zero centerline flow velocity. The parabolic VP gradually restores, but it takes 13.4 mm for the velocity to reach 90% of the maximal observed value (AD3 and AD4), while full insonation of the vessel is already reached after 6.49 mm (AD3) and 8.68 mm (AD4). In terms of PDS, the highest appearing frequency monotonically increases with increasing range gate and has not reached the theoretical maximal velocity after 20 mm. With DWoc, recovery of VP - a lowering of centerline velocity - takes >20 mm (AD3) and 15 mm (AD4). This effect, together with the asymmetrical vessel insonation leads to an overshoot of maximal frequency in PDS, followed by progressive decline towards the theoretical maximal value.

Conclusions: The presence of the DW has a significant impact on the VP and the derived PDS, and is likely to affect the accuracy of flow-estimation methods based on PDS analysis.

1128 Identification of nonparallel helical myofiber geometry of the left ventricular wall by high resolution B-mode ultrasound
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Background: Muscle fibers in the left ventricle (LV) have a non-parallel three-dimensional helical distribution that forms a right handed helix in the subendocardium and a left handed helix in the subepicardium. We hypothesized that variations in intensity of transmural ultrasound backscatter in excised and beating hearts using high resolution epicardial ultrasound will correlate with the transmural variations in myocardial fiber orientation.

Methods: The transmural ultrasound backscatter intensity and LV fiber orientation were compared in 14 porcine hearts (8 explanted and 6 beating hearts in situ) using 10 and 14 MHz linear array transducers. Results: Pannin cross-sectional view of LV in its short axis from apex towards base demonstrated a counter-clockwise rotation of the inner and outer layers, that correlated with an inverse helical arrangement of the epicardial and subendocardial regions. Reflections at the cardiac apex formed transmural arcs (Fig A) which corresponded with its spiral muscle fiber geometry. The mean difference in the end-diastolic acoustic intensity of subepicardial and subendocardial

High resolution ultrasound imaging

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