Ultrasonic Measurement of Tissue Motion Diagnostic Applications

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Ultrasonic examination is well suited for the measurement of tissue motions because of the superior temporal and displacement resolution. Temporal resolution of 0.1 milliseconds and displacement resolution of 100 nanometers are easily achievable with inexpensive, portable instruments. Normal motions and velocities and pathologic vibrations, velocities and expansions/compressions can be detected with ultrasound in all anatomic regions except the lung.

Ultrasonic Doppler methods for the detection and measurement of tissue velocity originated with Koneko and Satamura in 1957. Blood velocity was quantified by measuring the rate of phase change of the ultrasonic echo with time, which appeared as an audible frequency. Doppler velocimetry has become the standard for diagnosing cardiac and vascular stenoses, valvular regurgitation in the heart and veins and flow rates in named arteries and veins. Merrill Spencer and associates have pioneered applications of embolus detection in divers (bends), Patent Foramen Ovale (PFO) and cardiovascular surgery.

In addition to the measurement of blood velocity with ultrasound, ultrasound is capable of measuring the motion of solid tissues. In many ways, this is easier than measuring the speed of blood because the echo strength from solid tissues is 50 dB greater than the echo strength from blood. The observation of cardiac motions is the most common application. The simplest use of ultrasound to measure tissue motion is the M-mode echocardiogram. This B-mode technique is limited in displacement resolution to the depth resolution of B-mode ultrasound, which is the wavelength of ultrasound or about 300 micrometers. Doppler methods have a resolution at least 10 times better than M-mode resolution. In obstetrics, Doppler detected fetal cardiac motion detection has simplified the measurement of fetal heart rate. This method has been used for a quarter of a century. In modern cardiology, tissue Doppler methods and tissue strain rate imaging applied to the myocardium are vying for superiority in the detection of myocardial ischemia.

In 1969 Sumner, Hokanson and Strandness began exploring the motion of arterial walls as a measure of their mechanical properties using phase detection methods. Over the following 3 decades, tissue displacement resolution has improved from 100 micrometers to 40 nanometers. These methods are the basis of modern sono-elastic imaging methods. Sono-elastic imaging measures the mechanical properties of tissue by applying an external displacement or vibration to the tissues and observing variations in the tissue strain.

In our laboratory, we have used ultrasound echo phase detection methods to measure tissue motions induced by tissue patho-physiology. Six diagnostic methods will be discussed in this presentation: 1) vector Doppler velocimetry for the study of complex flow, 2) arterial wall motion for the detection of critical stenosis, 3) ultrasonic vibrometry for the detection of stenoses and internal bleeding, 4) differential tissue velocimetry for the detection of internal bleeding, 5) ultrasonic plethysmography for the detection of perfusion abnormalities including cancerous tumors, 6) transcranial ultrasonic brain velocimetry for the detection of intracranial bleeding.

Vector Doppler is a method that was developed to solve the problem of the Doppler angle during peripheral vascular examinations. In cardiac Doppler examinations of the aortic outflow track or the mitral

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inflow track, the ultrasound beam can be aligned with the axis of the vessel for flow measurements or with the axis of the jet in stenotic flow. The Doppler angle of examination is ZERO degrees for these measurements and the results can be used to correctly compute flow rates and Bernoulli pressure drops. The orientation of peripheral arteries and veins parallel to the skin prevents the transcutaneous ultrasound beam from intersecting the lumen at ZERO degrees to the vessel axis. Most examiners assume that the because the velocities must be parallel to the artery walls in simple Laminar flow, the velocities are parallel to the vessel axis. This allows the computation of an "angle adjusted velocity" by dividing the measured component of the velocity along the ultrasound beam direction by the cosine of the angle between the Doppler ultrasound beam and the vessel axis. Unfortunately, the assumption of para-axial velocities in peripheral vessels is not correct. In all peripheral arteries and veins, the laminar velocities are helical, parallel to the vessel walls but NOT parallel to the vessel axis. This assertion can be easily verified by measuring the "angle adjusted velocity" in any artery using a variety of achievable angles like 45 degrees and 60 degrees. Systematically, in all arteries except the distal superficial femoral artery, "angle adjusted" velocity measurements are higher at 60 degrees than an 45 degrees. This finding is exacerbated at the entry to a stenosis where the velocity vectors converge. Critics have denied this hemodynamic effect by suggesting that the finding is due to improper examination technique or instrument error. However the finding has been confirmed and found to be consistent in ultrasonic duplex scanners made by all manufacturers and examinations done by the most experienced and qualified examiners.

To allow the determination of correct para-axial velocity components for flow rate computations and correct intra-stenotic velocities for Bernoulli pressure depression computations in peripheral vascular examinations, we have constructed a Vector Doppler system. The system uses a single ultrasonic transmitter and multiple ultrasonic receivers to view different vector components of velocity simultaneously. The system is able to operate as a multigate Doppler along the transmitter beam pattern. Coherent eddies and flow rotations can be visualized using this method. For the study of post-stenotic eddies, which can have oscillation frequencies of several hundred cycles per second, cross-plane Doppler is allows direct measurement of the eddies and of the associated wall vibrations.

Ultrasonic Vibrometry is an imaging stethoscope. In the body, there are no normal extended sounds generated by the peripheral cardiovascular system. The normal short cardiac valve sounds are associated with valve closures. Longer duration heart murmurs and vascular bruits are always associated with some unusual or pathologic state. Systolic and diastolic heart murmurs and vascular bruits are markers of flows through energy dissipating orifices. The orifice can be a regurgitant or stenotic heart valve, a heart wall defect, a peripheral artery stenosis or a bleeding arterial puncture. The frequency of a murmur/bruit can be computed by using the Strouhal number (frequency * diameter/velocity) and the acoustic power of the bruit is equal to the pressure drop across the orifice times the volume flow rate. Lees and Dewey used the Strouhal number to predict that the diameter of the residual lumen in a carotid artery could be predicted by dividing the bruit frequency into the number 500 mm*Hz.

One of the difficulties in detecting bruits/murmurs is the short duration of the vibration. Wavelet analysis is ideally suited to the automatic detection of these vibrations. We have developed an ultrasonic system to image vibrations with frequencies up to 1 KHz and amplitudes as small as 40 nanometers. We have used the system to demonstrate peri-stenotic acoustic vibrations in arterial flow and vibrations in the vicinity of arterial bleeding.

Arterial Wall Motion measurements can be used to compute vascular compliance and to demonstrate normal arterial intimal function with changing flow rates (brachial artery resctivity). Another interesting application is to examine the motion of the arterial walls associated with a stenosis. In a normal artery the expansion during systole and relaxation during diastole can be observed, but when intra-stenotic velocities are high, the Bernoulli effect occuring systole can cause paradoxical pulsation in the stenotic region. This may be a marker for a critical arterial stenosis. In a patient with a normal 120/80 mmHg blood pressure, an arterial stenosis may cause systolic velocities of 400 cm/sec (with a Bernoulli

pressure depression of 64 mmHg) and diastolic velocities of 200 cm/sec sec (with a Bernoulli pressure depression of 16 mmHg). The stenosis region will have a luminal (venturi) blood pressure of (120-64) / (80-16) = 56/64 mmHg. The condition of lower intrastenotic blood pressure in systole than in diastole should cause the stenotic region of the artery to exhibit paradoxical pulsation, smaller in systole than in diastole.

Ultrasonic Plethysmography is a technique based on traditional volume plethysmography and photo-plethysmography. Typically, all peripheral tissues expand about 0.1% by cross section during each systole. Using ultrasound, an image space can be divided into a series of voxels with 1 mm depth dimensions. By measuring the motion of the boundaries of such voxels with a displacement resolution of 0.1 micrometers or less, the expansion of each voxel can be measured with a resolution of 0.01% or 10% of the expected pulse amplitude.

Using this technique, we are seeking to identify the increased pulsatility associated with angiogenesis around cancerous tumors. We have succeeded in the differentiation of some tissue types. The subcutaneous fat in breast has a pulse amplitude of about 0.02%. in addition to measuring the pulse amplitude at the heart rate, we also measure the phase and amplitude of higher harmonic components of the expansion. The phase of the pulse amplitudes has demonstrated a surprising finding: while some voxels exhibit pulse amplitudes of 0.7%, others exhibit an inverted pulse amplitude. The physiological explanation of this finding is related to the mechanical constraints on tissue.

Differential Tissue Velocimetry is a technique that looks at the measured derivative of the depth component of tissue displacement with depth. Solid tissues undergoing expansion, compression or contraction are expected to have continuous derivatives with depth. The derivative can be called tissue strain. Discontinuities in the depth derivative marks regions of "slip planes" that indicate tissue boundaries.

Normal tissue boundaries, such as the pleura or the peritoneum exhibit high discontinuous strain with both the respiratory and the cardiac cycle. Generally the respiratory component is greater than the cardiac component. in addition, the strain is cyclic with both the respiratory and the cardiac cycle. In contrast, strain discontinuities associate with organ fracture do are not restored to ZERO at the end of the respiratory or cardiac cycle, but may expand for a long period as the fracture fills with blood and then empty into the peritoneum.

Transcranial Brain Velocimetry is under development for the detection of intracranial bleeding. Alternatives for the detection of intracranial bleeding require expensive CT or MR images of the brain. In a typical intra-cranial bleed, the hematoma will expand at a slow rate over half a day to displace the cerebral midline by 10 mm. On the average, the sideways velocity of the brain midline is 1 mm/hr. There is no assurance that the bleed is continuous and uniform over the period of the bleed. Traditional brain imaging detects the displacement of the midline and the presence of a large hematoma.

Ultrasonic trans-cranial Doppler methods can detect the 1 mm/hour lateral velocity expected during unilateral intracranial bleeding. These methods can also detect other motions of the brain associated with the cardiac cycle, the respiratory cycle and with gravitation and accelerational motions of the skull. In addition to the possible application of this method for the study of intracranial bleeding, studies of brain motion during trauma may be intriguing.

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