**ECG-GATED, MECHANICAL AND ELECTROMECHANICAL WAVE IMAGING OF CARDIOVASCULAR TISSUES IN VIVO**

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Abstract—In simplistic terms, the motion of the heart can be summarized as an active contraction and passive relaxation of the myocardium. However, the local motion of cardiovascular tissues over the course of an entire cardiac cycle results from various transient events such as the valves closing/opening, sudden changes in blood pressure and electrical conduction of the myocardium. The transient motion generated by most of these events occurs within a very short time (on the order of 1 ms) and cannot be imaged correctly with conventional imaging systems, due to their limited temporal resolution. In this paper, we propose a method for imaging this rapid transient motion of tissues in cardiovascular applications. Our method is based on imaging tissues with ultrasound at high frame rates (up to 8000 fps) by synchronizing the two-dimensional (2D) image acquisition on the electrocardiogram (ECG) signals. **In vivo** feasibility is demonstrated in anesthetized mice. The propagation of several transient mechanical waves was imaged in different regions of the myocardium and the wave phase velocities were found to be between 0.44 m/s and 5 m/s. These waves may be generated by either a purely mechanical effects or through electromechanical coupling in the myocardium depending on the phase of the cardiac cycle, in which they occur. The abdominal aorta was also imaged using the same technique and the propagation of a mechanical pulse wave was imaged. The pulse wave velocity was measured and the Young’s modulus of the vessel wall was derived based on the Moens-Korteweg equation. This method could potentially be used for mapping the stiffness of the myocardium and the artery walls and may lead to the early diagnosis of cardiovascular diseases. (E-mail: ek2191@columbia.edu) © 2007 World Federation for Ultrasound in Medicine & Biology.

Key Words: Cardiac, Contraction, Displacement, ECG gating, Electromechanical, High-frequency, High resolution, Mechanical, Mice, Motion, Myocardium, Ultrasound.

**INTRODUCTION**

The evaluation of the heart’s function using conventional imaging techniques is currently based on the mechanical interpretation of the main task of the heart: this is pumping blood through an active contraction phase (systole) and a passive relaxation phase (diastole) of the myocardium. To quantify the mechanical properties of the myocardium, several techniques have been introduced, such as tissue Doppler imaging (Sutherland et al. 1995), strain rate imaging (Heimdal et al. 1998), elastography (Konofagou et al. 2002) in the field of ultrasound imaging or cardiac tagging (Zerhouni et al. 1988; Declerck et al. 2000) in the field of magnetic resonance imaging. These techniques allow the measurement of the myocardial deformations over a complete cardiac cycle, but, due to their temporal resolution (5 ms at best), they remain limited to the global motion of the heart.

However, in addition to the slow and large component of the heart beat, some transient mechanical vibrations are constantly generated in the heart and the arteries. Since the early 19th century, it has been established that the aortic valve closing or opening generates a transient pressure wave in the arteries, known as the arterial pulse wave (Young 1809). This pressure wave propagates in blood vessels at high speed (on the order of 1 to 10 m/s) and, to our knowledge, the pulse wave propagation in arteries has never been imaged or visualized by conventional imaging techniques, due to their relatively low temporal resolution. To overcome this limitation, noninvasive techniques have been developed to measure the time that it takes for the arterial waveform...
to traverse two distant points of the artery, using Doppler ultrasound (Avolio et al. 1983) or pressure sensors affixed onto the skin surface (McLaughlin et al. 2003). The velocity of the pulse wave has been widely investigated in vivo and it is well known that the average pulse wave velocity is related to the compliance of the vessel wall (Greenwald 2002). A significant compliance increase has been reported with age (Rogers et al. 2001) and with atherosclerosis (Wang et al. 2000). Moreover, further evidence of rapid transient motion in cardiovascular tissues was found by Kanai et al. (1993; 2001) inside the heart. They found that a transient mechanical wave was generated at the root of the aortic valve, due to a strong mechanical effect upon valve closure. The pulse wave was shown to propagate across the intraventricular septum at speeds that ranged between 1 m/s and 5 m/s.

Kanai et al. (2005) also established a model to derive the viscoelastic parameters of the myocardium from the wave velocity.

To obtain a spatiotemporal image of such mechanical waves and to measure the wave velocity, the motion of tissues must be imaged at high frame rates. Kanai et al. (1993) developed a customized ultrasound system that used a small number of ultrasound beams distributed over a large field-of-view. Using this technique, they were able to simultaneously measure the tissue motion in the myocardium using 16 ultrasound beams at a high frame rate (450 Hz). However, due to the limited number of beams, the image quality was compromised and motion could only be estimated at a small number of predetermined locations. Using the same technique, they also measured the regional pulse wave velocity in carotid arteries (Kanai et al. 2000) with four ultrasound beams. Another approach has been developed by others (Sandrin et al. 1999; Tanter et al. 2002) that uses original beamforming emission sequences for imaging the shear wave propagation in soft tissues. They developed an ultrafast scanner that can achieve two-dimensional (2D) real-time acquisitions at a frame rate of 5000 fps. However, this technique reduces the spatial resolution and requires complicated and costly ultrasound imaging equipment.

To achieve high frame rate ultrasound imaging, another approach based on the use of electrocardiogram (ECG)-gated data acquisition has been recently proposed for cardiac and vascular applications (Pernet and Konofagou 2005; Williams et al. 2005; Li et al. 2005; Cherin et al. 2006). Based on the same principle, we propose here a new method for imaging the mechanical vibrations that propagate in cardiovascular tissues. Taking advantage of the ECG, which is a stable and accurate indicator of the cardiac cycle, our method is based on imaging the tissue with high pulse repetition frequency (PRF), high resolution and synchronizing the 2D frame acquisition with the ECG over several hundreds of cardiac cycles. Although the 2D acquisition is not performed in real-time, this method combines the advantages of having a very high frame rate of up to 8000 fps, a high spatial resolution and a full 2D view of the heart.

Using a speckle-tracking motion estimation technique based on the radio-frequency (RF)-signals, small displacements of the tissue on the order of 1 μm were detected between consecutive frames. The 2D local displacements maps of the left ventricular wall of a mouse were processed for the entire cardiac cycle at high frame rates and those showed the main components of the myocardial motion as well as several transient and rapid motions. These transient motions were analyzed in the frequency range of 50 to 500 Hz. The analysis showed the propagation of several mechanical waves and their wave speeds were measured. The same method was also applied to the abdominal aorta to image the arterial pulse wave with a high frame rate and to determine the local pulse wave velocity with great precision. The motion of the aortic wall was captured at 8000 fps and our results showed the propagation of a strong pulse wave due to the sudden pressure change in the aorta during the cardiac cycle. The pulse wave velocity was measured and the local elasticity of the vessel was derived. This method could potentially be used for mapping the stiffness of the myocardial and the arterial walls and may lead to the early detection and reliable diagnosis of cardiovascular diseases.

**MATERIALS AND METHODS**

**Animal preparation**

The mice were anesthetized with tribromoethanol. The hair was removed using potassium thioglycolate and the mouse was placed in the supine position on a heating stage (VisualSonics, Toronto, ON, Canada) to keep the body temperature steady. The ECG signal was obtained from the extremities. The ultrasound probe was placed on the chest or the abdominal wall using degassed ultrasound gel (Aquasonic 100, Parker Laboratories Inc., Fairfield NJ, USA) as a coupling medium. All animal experiments complied with legal requirements and institution guidelines according to the Columbia University Institutional Animal Care and Use Committee.

**RF signal acquisition**

An ultrasound scanner specifically developed for imaging small animals (Vevo 770, Visualsonics, Toronto, ON, Canada) was used for this study. The high-frequency ultrasound probe was composed of a single-element focused transducer working at 30 MHz, with a focal depth of 12.7 mm. The transducer moved laterally with a very slight arc and real-time 2D images could be
acquired at a frame rate of up to 60 Hz. The field-of-view was 12×12 mm, the axial resolution was 50 μm, and the lateral resolution was 100 μm.

A digitizer (two channels, 200 MS/s, 14 bits, CS14200, Gage Applied Technologies, Lachine, QC, Canada) mounted on a PC computer slot was connected to the analog RF-output of the ultrasound scanner.

The ultrasound probe was placed on the chest in the parasternal position to obtain a longitudinal (long-axis) view of the left ventricle of the heart. The probe could also be positioned over the abdomen to obtain a longitudinal view of the abdominal aorta.

**High frame rate acquisition**

In addition to the real-time scanning mode, a high frame rate acquisition mode (EKV™ or, ECG-based Kilohertz Visualization) was provided on the scanner to allow detailed visualization of the heart contraction. Using this technique, the ultrasound acquisition of each RF-line was triggered on the mouse ECG. The transducer was moved laterally and, for each position of the transducer, ultrasound echo signals were recorded with a PRF of 8000 pulses/s during several cardiac cycles. The ECG was simultaneously recorded and thus allowed for the synchronization of the RF-lines based on the R-wave peak (a reliable peak of the ECG during the cardiac cycle). The complete 2D acquisition duration was approximately 5 min.

To be able to compute the tissue motion with this mode, RF signals and ECG signals were digitized during the EKV acquisition and transferred to the computer in real-time. The data were then processed off-line: RF-lines were synchronized using the R-wave peak of the ECG signal and a complete set of 2D ultrasound RF-data were reconstructed at 8000 fps for one complete cardiac cycle (approximately 150 ms).

**Motion estimation**

The motion of the tissue was estimated off-line using an established classical speckle-tracking method (Bonnefous and Pesque 1986). This technique was based on detecting the small local displacements of the tissue that occur between two consecutive frames. With the current method, only axial displacements (i.e., along the axis of the transducer beam), which coincided with the radial displacements in a long-axis view, were estimated. Esti-
mation of lateral displacement was not deemed necessary for correction of decorrelation, due to the very high frame rate used. Therefore, such a study goes beyond the scope of this paper, but will be addressed at greater depth in future investigations. In our algorithm, the time shifts in the backscattered signals were determined between the two consecutive frames through cross-correlation of small sliding windows over the entire RF-line. This technique allowed the detection of very small displacements on the order of 0.1 μm or less (correlation windows of 150 μm, overlapping 90%). Finally, the ciné-loop of the axial displacements was generated at a frame rate up to 8000 frames/s for the entire cardiac cycle.

Frequency analysis

Tissue axial displacements were estimated as a function of time at a fixed depth. A Blackman window of 100 points corresponding to a duration of 25 ms was moved along the displacement variation at a fixed depth, by steps of 2 ms. The windowed signals were zero-padded to 8192 points and their Fast Fourier Transform (FFT) was calculated. The frequency content of the displacements was evaluated graphically by plotting these spectra as a function of time. Based on this frequency analysis, the transient and slow motions of the tissues were separated using a digital filter. The displacement estimates were temporally filtered using an FIR band-pass filter with cut-off frequencies of $f_1 = 50$ Hz and $f_2 = 500$ Hz, which removed high-frequency noise in addition to the low-frequency components.

Pulse-wave velocity

To analyze the propagation of the mechanical waves, the phase velocity of the vibration was determined for particular angular frequencies $\omega$. The wave was assumed to propagate with a velocity $c$ in a direction $r$ that was arbitrarily determined on the image by the direction of the wall and a set of measurement points was selected on this direction. The wave number is $k = \omega c$ and the phase of the wave is $\varphi(r) = kr$ along the direction of propagation. The phase was measured as a function of the propagation distance $r$, using the Fourier transform of the temporal displacements at the location $r$ computed at the angular frequency $\omega$. Finally, the derivative of the phase of the wave with respect to distance was estimated using a linear regression fit on the set of measurements points and the velocity of the wave at the frequency $f$ was calculated:

$$c(f) = \frac{2\pi f}{\partial \varphi/\partial r}. \quad (1)$$

Modulus estimation

The theory of elastic wave propagation in soft biological tissue was considered to derive the Young’s mod-

![Fig. 3](image-url)

Fig. 3. (a) Temporal variation of the axial displacements estimated on one central RF-line (shown as the white vertical line plotted on Fig. 2b). The motion of the wall exhibits two types of variations: these are slow and large variations during the systolic and diastolic phases and small and rapid variations between the two main phases. (b) Frequency content of the displacement variation in the septum at a fixed depth of 12.5 mm plotted as a function of time. The depth location is shown by a horizontal dash line and the size of the Blackman window used for the frequency analysis is indicated by brackets in (a). (c) Temporal variation of the axial displacements after bandpass filtering showing the transient and high-frequency components. (d) ECG signal acquired simultaneously.
fluid is well described by the Moens-Korteweg equation (Korteweg 1879; Moens 1879):

\[ c = \sqrt{\frac{Eh}{2R\rho}} \]  

(2)

where \( c \) is the velocity of the wave, \( E \) is the Young’s modulus of the conduit wall, \( h \) is the wall thickness, \( \rho \) is the density of the wall (similar to the density of the fluid) and \( R \) is the radius of the tube. According to this equation, the elasticity of the vessel wall can be derived from the measurement of the pulse wave velocity in the artery.

**RESULTS**

*In vivo cardiac imaging*

Figure 1 shows a B-mode image of a typical parasternal long-axis view obtained in a normal mouse. This image shows the main structures of the left ventricle: these are the interventricular septum, the cavity of the left ventricle, the papillary muscle and the posterior wall, and the outer boundary, which is visible due to strong reflections at the epicardium-lung interface. In this experiment, the average duration of a full cardiac cycle was 138 ms. Axial displacements were estimated for the complete set of data. To keep the displacements at optimal magnitudes for the estimation (on the order of 0.1 \( \mu m \); Walker and Trahey 1995) and to reduce the amount of data, the number of frames acquired was reduced by a factor of two, which also reduced the frame rate to 4000 fps. Figure 2 shows the color-coded axial displacements overlaid onto the gray-scale B-mode image for two different phases of the cardiac cycle. During the systolic phase, the contraction of the myocardium is shown through positive displacements (red) of the posterior wall.

![Sequence of images around end-systole every 0.6 ms showing the propagation of a mechanical wave in the septum (from right (base) to left (apex)). The wavelength of the mechanical vibration is on the same order as that of the image lateral dimension and only a small part of the oscillation can be seen on each image. The white arrows indicate the progression of the wavefront in (a), (b), (c) and (d) followed by (e) and (f), where a second wavefront is shown by the yellow arrows.](image-url)
and negative displacements of the septum (blue) (Fig. 2a), whereas, in the diastolic phase, the directions of the displacements (and the colors) are reversed during the beginning of systole (Fig. 2b). It should be noted that, even if a large part of the myocardium of the posterior wall is not visible, the motion of the epicardium undergoes similar motion.

A temporal analysis of the motion was performed for single RF lines of the image. Fig. 3a shows tissue axial displacements along the central line of the image, indicated by the white dotted vertical line on Fig. 2b, as a function of time (horizontal direction) and depth (vertical direction). Fig. 3d shows the ECG signal that was measured during this acquisition. In Fig. 3a, the motion of the septum, the papillary muscle and the posterior wall are shown in color over two cardiac cycles. Motions obtained from tissue moving toward the transducer (at the top of the image) are displayed in red and those moving away, in blue. It shows the successive main phases of the cardiac cycle: these are the contraction of the myocardium (systole) initiated at the R-wave peak of the ECG, followed by the relaxation phase (diastole). The duration of the active contraction was approximately 50 ms and that of the relaxation was 35 ms. In addition to this slow and large motion, some rapid transient variations of a few ms were observed at the beginning and at the end of the systolic phase, in the septum and the posterior wall.

The frequency content of tissue displacements in the septum (at a fixed depth of 12.5 mm) was analyzed between 0 Hz and 500 Hz and is shown as a function of time in Fig. 3b. During the contraction and the relaxation of the heart, the motion of tissue was found to be in the low-frequency range of up to 60 Hz. However, during the transient motion at the end of systole, much larger frequency components were found that ranged between 50 Hz and 500 Hz. The same effect was found for the transient motion at the beginning of systole, but the frequency range was limited to between 50 Hz and 250 Hz. Thus, it was possible to almost completely separate the transient part of the displacement by filtering out the low-frequency component of the motion. After filtering the displacements using a Finite Impulse Response (FIR) band-pass filter with cut-off frequencies of \( f_1 = 50 \) Hz and \( f_2 = 500 \) Hz, the two vibrations were clearly visible and are shown on Fig. 3c. These rapid variations occurred within less than 3 ms around the beginning of systole and end-systole (see arrows in Fig. 3).

**End of systole**

To spatially analyze the vibration around end-systole, we considered the data between 52 ms and 70 ms after the peak of the R-wave. Fig. 4 shows a sequence of axial displacements overlaid onto the gray-scale B-mode images at every 0.6 ms around end-systole. This sequence uncovers a strong mechanical wave propagating in the longitudinal direction of the ventricle along the myocardium, from the base (right side of the images) to the apex (left side). In other words, as the tissue locally vibrates along the axial direction of the beam (i.e., along the beam axis), a transverse wave propagates along the lateral direction (i.e., in-plane, perpendicular to the beam axis).

The mechanical wave was visible in both the posterior wall and the septum, but its amplitude was eight times higher in the septum. Therefore, in this study, only the wave propagating in the septum was considered. A set of 60 samples was selected over consecutive RF-lines in the septum along the propagation direction (lateral direction of the image) and the phase of the wave was computed at different frequencies. Three frequencies were selected for which the displacement amplitude was large enough: these were 82 Hz, 246 Hz and 410 Hz. The phase velocity of the wave was computed for these frequencies and a large dispersion was found. The propagation was plotted in Fig. 5 as a function of the phase of the wave divided by the angular frequency. The phase velocity was found to be 1.20 m/s at 82 Hz, 3.02 m/s at 246 Hz and 4.21 m/s at 410 Hz. Note that the cause for the step-like appearance of the curves (Fig. 5) was not found, but was not due to the phase unwrapping technique, as verified by appropriate methods.

**Beginning of systole**

The same analysis was performed at the beginning of systole. The filtered data were processed between 0 ms and 20 ms from the peak of the R-wave. Fig. 6 shows a
sequence of axial displacements overlaid onto the gray-scale B-mode images every 2.8 ms around the beginning of systole. A strong vibration was found in the septum, but no evidence of wave propagation was found in the image plane. Therefore, a mechanical wave may propagate in the perpendicular direction, but could not be observed in the 2D image plane.

However, Fig. 6 does depict a wave propagating in the posterior wall (see the white arrows). The displacements were initiated at the apex (left side of the images) and then propagated toward the base (right side). The phase velocity was determined at different frequencies using the method previously described. However, this vibration had a much smaller bandwidth than that of the end-of-systole vibration. Fig. 3b shows that the frequencies ranged only up to 230 Hz. In this bandwidth, we measured a constant velocity without any significant change as a function of the frequency. The distance of propagation is plotted in Fig. 7 as a function of the phase of the wave divided by the angular frequency. The phase velocity of the wave was obtained using a linear regression fit and was estimated to be 0.44 m/s.

In vivo vascular imaging

A longitudinal view of the abdominal aorta of a mouse was imaged using the high frame rate technique. Axial displacements were calculated and the movie of the motion was processed at 8000 fps for a complete cardiac cycle. During the cardiac cycle, the displacements of the arterial wall were found to be very small except after the beginning of systole. Strong displacements of the wall started 10.3 ms after the R-wave peak of the ECG. Fig. 8 shows a sequence of the axial displacements in color overlaid onto the gray-scale B-mode image. A transverse wave started propagating on the right side of the images (heart side) and then propagated toward the left side in less than 3 ms. This transverse wave was generated from the sudden pressure change of
the blood bolus traveling through the vessel, known as the arterial pulse wave (Nichols and O’Rourke 1998).

The phase velocity of the pulse wave was computed at the frequency of 200 Hz. The distance of propagation is plotted in Fig. 9 as a function of the phase of the wave divided by the angular frequency; the phase velocity was obtained using a linear regression fit and was found to be 3.08 m/s. The radius of the vessel \( r = 0.47 \) mm and the wall thickness \( h = 0.12 \) mm were estimated from the B-mode images and the blood density was assumed to be 1060 kg/m\(^3\) (Cutnell and Kenneth 1998). Using these parameters, the Young’s modulus of the aortic wall \( E = 78.8 \) kPa was derived from the Moens-Korteweg equation (eqn 2), which is similar to previous reports for thoracic aortic moduli in biomechanics literature (Fung 1993).

**DISCUSSION AND CONCLUSION**

The heart and the aorta of mice were imaged in vivo using a high-frequency ultrasound 2D scanner. A technique was developed to image the motion of cardiovascular tissues with frame rates up to 8000 fps, by synchronizing the acquisition on the ECG signal. The technique of ECG-triggering has been used recently by several groups (Pernot and Konofagou 2005; Li et al. 2005; Cherin et al. 2006), but this was the first time, to our knowledge, that mechanical and electromechanical wave propagation as well as rapid vibrations could be visualized in the heart muscle and aorta using ECG-gated techniques, i.e., at simultaneous high frame rate, high spatial resolution, high precision in the motion estimation and full 2D mapping. Thus, this ultrafast imaging method revealed that several transverse mechanical waves were generated during the cardiac cycle in the heart and the aorta and propagated in tissues at speeds ranging between 0.4 m/s and 5 m/s.

The propagation of low-frequency transverse waves is closely dependent on the viscoelastic properties of the tissue and therefore the propagation of shear waves in soft tissues has been widely investigated for elasticity estimation, using ultrasound imaging (Sarvazyan et al. 1998; Sandrin et al. 1999) or MRI (Muthupillai et al. 1995; Sinkus et al. 2000). Several methods have been proposed to generate low-frequency shear waves but all these techniques require generating shear waves artificially, using an external vibrator (Muthupillai et al. 1995; Sandrin et al. 1999; Sinkus et al. 2000) or the acoustic radiation force as an internal volumic force (Sarvazyan et al. 1998; Nightingale et al. 2003; Bercoff et al. 2004).

In this paper, we proposed to image another kind of mechanical wave: this is the natural and spontaneous wave generated by the heart that propagates in soft tissues. First, we found that strong transverse mechanical waves are spontaneously generated during the cardiac cycle in the myocardium and the aortic wall. Therefore, contrary to other organs that require generating artificially mechanical waves, the cardiac pulsation induces its own natural waves that can be used for the estimation of the viscoelastic properties. Second, the ECG signal can be used to synchronize multiple ultrafast one-dimensional ultrasound acquisitions over hundreds of cardiac cycles, enabling the mapping of mechanical waves propagation at very high frame rates.

Although this technique is not in real-time, the relatively slow and large motions induced by respiration did not pose a problem in the displacement estimation, because of the high effective frame rate: in our experiments, axial displacements were estimated on individual RF lines between two sets of data acquired within 0.25 ms. However, in theory, respiration and other large motions would affect the 2D spatial reconstruction to a larger extent and could induce significant shifts between the lines of the B-mode and the displacement images. In spite of this, our results showed that motion artifacts remained very small with mice deeply anesthetized under physiological respiration, even for acquisitions of more than 2000 cardiac cycles (i.e., approximately 4 min). Sophisticated respiratory gating could be added to the acquisition system completely to remove these artifacts.

Another potential issue in the ECG-gated acquisition technique is the effect of strong irregularities of the cardiac cycle. In case of arrhythmia, for example, the
line-by-line-based acquisition could be severely desynchronized and may introduce artifacts in the 2D reconstruction of the motion.

The wave propagation could potentially be used to map the viscoelastic parameters of the myocardium and the vessels, which is very important for the diagnosis of cardiovascular diseases such as coronary artery disease or atherosclerosis. However, to become clinically useful, the origin of these mechanical waves must be better understood. Although the arterial pulse wave is relatively well known, the origin and the propagation of mechanical waves in the myocardium has not been well described and may depend strongly on the phase of the cardiac cycle. Kanai et al. (2001) observed a strong mechanical wave at end-systole in the human heart and they demonstrated that this wave was generated at the same time as the valve closure of the aortic valve. They concluded that it was a purely mechanical wave induced by the strong vibration upon valve closure and they recently modeled the wave propagation in the interventricular septum as a Lamb wave guided by the two interfaces of the left and right ventricles (Kanai 2005).

Fig. 8. Sequence of images showing the propagation of the pulse wave in the aorta every 0.7 ms. The ECG signal is plotted below each image and the red point indicates the time of the acquisition during the cardiac cycle.

At the beginning of systole, however, our results suggest that the wave propagation may be of a completely different nature. We showed that a strong me-
the myocardium (Konofagou et al. 2006) and its correspondingly induced by the electromechanical coupling mechanism of the electrical activation of the left ventricle effect at the base side. Therefore, in this case, the possibility of being generated by a valve closing/opening at the apex and then propagated toward the base, which is the opposite direction than the end-systolic wave propagation. This direction of propagation excluded the possibility of being generated by a valve closing/opening effect at the base side. Therefore, in this case, the mechanical wave appeared to be associated with the propagation of the electrical activation of the left ventricle and induced by the electromechanical coupling mechanism during electrical propagation. Ongoing investigations deal with measurements of electrical conduction in the myocardium (Konofagou et al. 2006) and its correlation with the mechanical wave propagation and clinical applications of the methods described.

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