Real-Time System for Respiratory-Cardiac Gating in Positron Tomography

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Abstract

A Macintosh-based signal processing system has been developed to support simultaneous respiratory and cardiac gating on the ECAT EXACT HR PET scanner. Using the LabView real-time software environment, the system reads analog inputs from a pneumatic respiratory bellows and an ECG monitor to compute an appropriate histogram memory location for the PET data. Respiratory state is determined by the bellows signal amplitude; cardiac state is based on the time since the last R-wave. These two states are used in a 2D lookup table to determine a combined respiratory-cardiac state. A 4-bit address encoding the selected histogram is directed from the system to the ECAT scanner, which dynamically switches the destination of tomograph events as respiratory-cardiac state changes. To test the switching efficiency of the combined Macintosh/ECAT system, a rotating emission phantom was built. Acquisitions with 25 msec states while the phantom was rotating at 240 rpm demonstrate the system could effectively stop motion at this rate, with approximately 5 msec switching time between states.

I. Introduction

Respiratory-cardiac gated Positron Emission Tomography (PET) acquisitions require a real-time system which translates analog inputs from respiratory and cardiac sensors into a digital output state. For doubly-gated acquisitions on the CTI/Siemens ECAT EXACT HR PET scanner, the system must select one of sixteen 47-plane histograms in scanner memory during any particular phase of the cardiac and respiratory cycles. Tomograph events are then redirected by the scanner to the selected histogram as the respiration and cardiac cycles progress.

Prior papers have reported algorithms for processing data that we have obtained via our doubly-gated acquisition system [1, 2, 3, 4, 5]. The focus of this paper is the implementation and performance of the Macintosh-based signal processing system which carries out the front-end tasks, and a performance characterization of the switching efficiency for the combined Macintosh/ECAT system.

The Macintosh system is interfaced to the ACS I version of the control system for the ECAT scanner. Software in the scanner has been modified to respond to state changes from the Macintosh and thus support doubly-gated acquisitions. Onboard the Macintosh, a National Instruments data acquisition board is combined with the real-time LabView software environment to provide a versatile platform which is easy to program and operate.

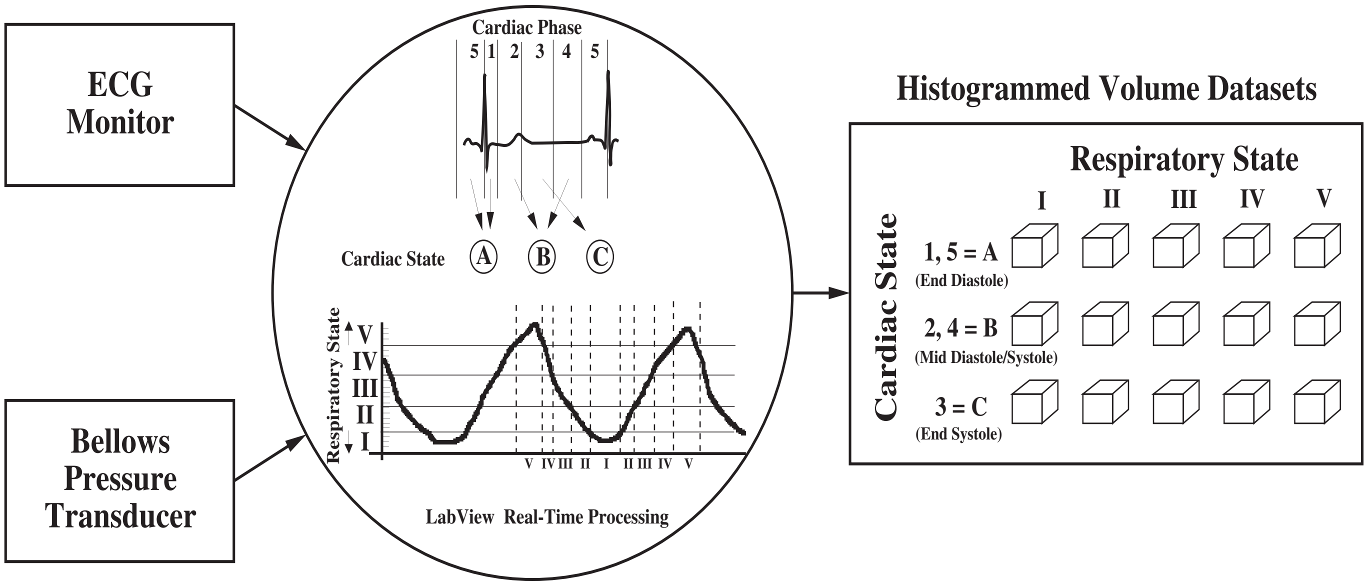

Figure 1: Schematic of doubly-gated PET acquisition. Real-time data acquisition hardware located in a Macintosh workstation processes respiratory and ECG inputs to supply the ECAT scanner with a suitable gate address. The net result of a gated acquisition is a 2D array of reconstructed image volumes. In this example, five cardiac phases are used to select three cardiac states. The respiratory cycle is divided into five states. These volumes are subsequently registered to each other and combined to form a composite PET image with reduced motion blur and without loss of statistics.

II. DOUBLE GATING

A double gating scheme for acquisition and subsequent processing of cardiac PET images is motivated by the assumption that the heart moves independently with two types of motion during a PET acquisition. Contractile motion associated with the cardiac cycle changes primarily the *shape* of the heart. Respiratory motion changes primarily the *position and orientation* of the heart as the diaphragm and chest wall move with each breath. The gating scheme described here separately bins PET events into a 2D array of histograms, where one axis is the state of the cardiac cycle, and the other is the state of the respiratory cycle (Figure 1). Respiratory and cardiac motion may then be captured in a series of freeze frames as each histogram is separately reconstructed into an image volume. Unfortunately, as the total number of tomograph events is distributed into many image volumes, the image quality of each reconstruction suffers due to lack of statistics. The image quality can be improved by adding the data back together, but only after the image volumes have been registered to each other. This registration task is a formidable problem, since it requires estimation of a motion field that matches cardiac features on a voxel-by-voxel basis from one reference dataset to corresponding features in all the other reconstructed volumes. Preliminary results have been presented which indicate that respiratory and cardiac motion can be estimated from routinely acquired doubly gated studies on dogs [4] and humans [1, 3], so it appears that motion compensation in a doubly-gated study is feasible.

Because the difference in the cardiac image between two reconstructed respiratory states is primarily a rigid translation and rotation, a simple rigid-body registration is used to add corresponding voxels in these states [1, 6]. To add corresponding voxels from different cardiac states, the volume must be nonrigidly deformed and added[3, 4]. Ideally, all states are registered to a reference volume and summed to produce a single composite image volume. Since all the data are added, the composite volume uses all the statistics available from the gated PET study, yet because the image data have been warped to match the shape of the heart at a reference position (typically end diastole at end expiration) there is little motion blur.

Detection of an R-wave from standard electrocardiograph (ECG) signals is used to monitor the phase of the cardiac cycle, similar to previous cardiac gating efforts[7, 8, 9, 10]. For monitoring the respiratory cycle, we use a pneumatic bellows secured around the patient's chest. Originally designed for use with MRI scanners, the pneumatic bellows attaches to a pressure transducer and generates an analog voltage related to chest expansion. A raw waveform from the bellows obtained during a typical study is seen in Figure 2. Close examination of the waveform reveals that it is relatively stable, yet sensitive enough to even detect small pressure fluctuations due to the heart beat (see Figure 2 inset).

III. HARDWARE / SOFTWARE

An overall schematic for the double gated system is seen in Figure 1. Analog signals from the ECG and respiratory monitors are directed to a Macintosh IIci computer, where they are

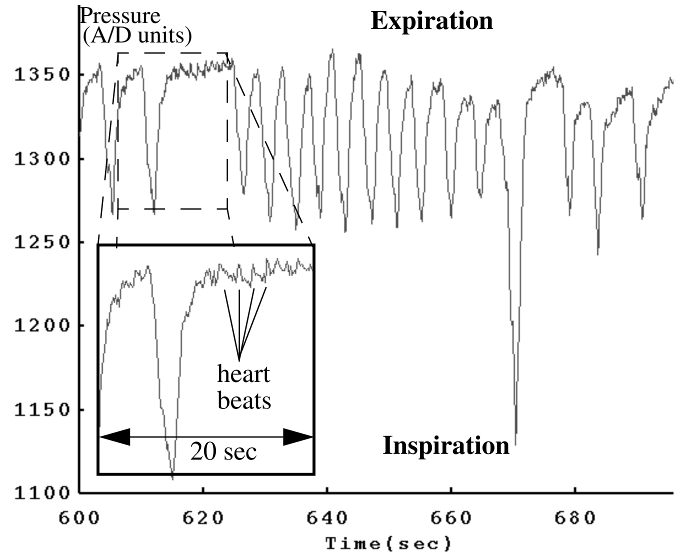

Figure 2: Raw digitized waveform from the respiratory bellows pressure transducer. Absolute waveform values are used to determine the respiratory state.

digitized and processed by LabView software to determine the current cardiac and respiratory states. The two input states are used to determine a 4-bit output state sent to the ECAT scanner. Each change in the output state initiates a reconfiguration in the scanner so that it directs tomograph events to the selected one of up to 16 separate histograms in scanner memory.

A more detailed look at the processing inside the Macintosh is shown in Figure 3. All input and output is achieved using a National Instruments NB-MIO-16 data acquisition board resident in the Macintosh. Control of this board and processing of the signals is carried out using the LabView software environment. The LabView environment allows considerable flexibility in processing data, yet allows real-time control of input and output signals.

To determine the cardiac state, the ECG signal is sent to processing hardware which generates a pulse at each R-wave. The R-wave pulse resets a continuously running hardware counter/timer onboard the data acquisition board. A tight loop in the control software frequently samples the value of the counter, thus obtaining the current time elapsed since the last detected R-wave. The cardiac gating phase is set directly using this elapsed time value. Note that different cardiac *phases* can be combined into the same cardiac *state*. For example, we combine the filling and emptying phases on either side of end systole into a single "transition" cardiac state. Similarly, the cardiac phase immediately after the R-wave is combined with data at the end of the cardiac cycle into a single "end-diastole" state.

An A/D converter samples the respiratory bellows transducer output, and the amplitude of the sample is used to set a respiratory state. We divide the range of expected respiratory sample values into equally spaced intervals whose number equals the number of chosen respiratory states. The respiratory state is therefore determined directly by the interval into which the sample value falls. Typically, respiratory sampling is done

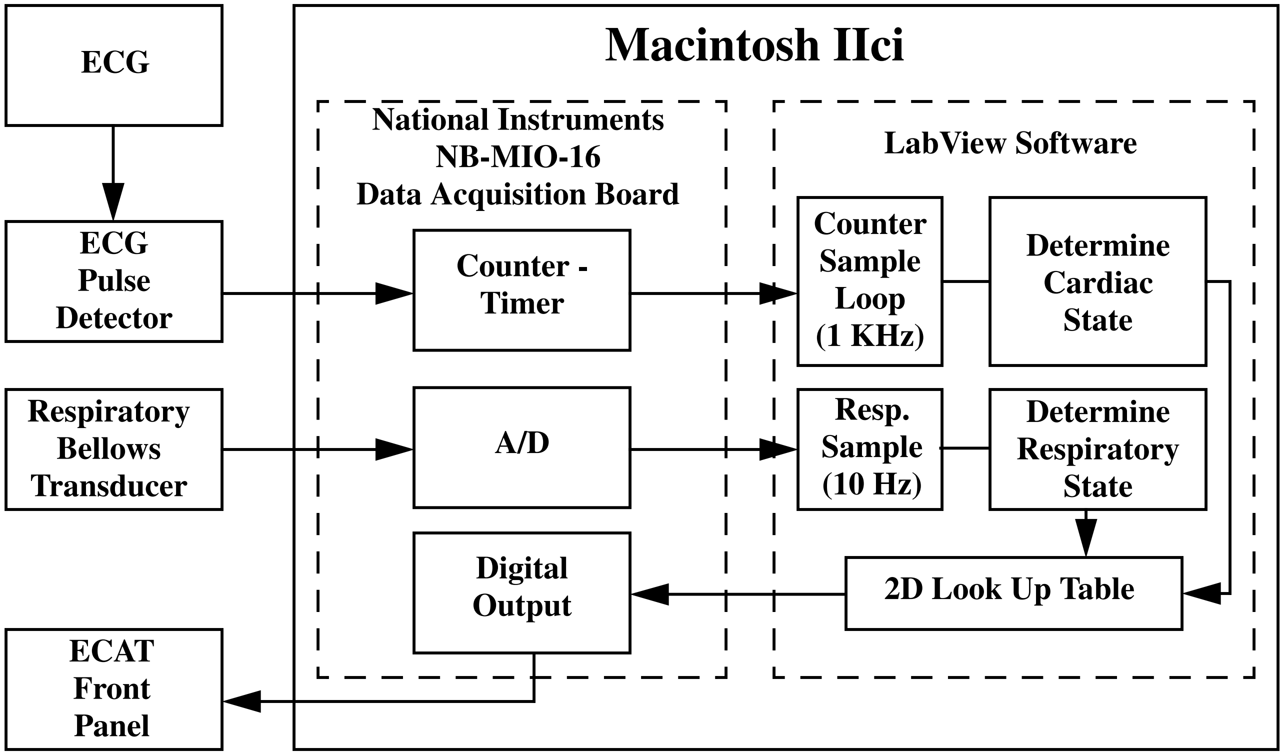

Figure 3: Data processing flow diagram. Analog signals from the ECG and respiratory bellows are directed to a National Instruments data acquisition board resident in a Macintosh IIci computer. An ECG-derived R-wave pulse resets a counter timer which is sampled at 1 KHz to determine the cardiac state. The respiratory waveform is digitized and used to determine a respiratory state. Cardiac and respiratory states select a 4-bit output state from a 2D lookup table. This output state is directed to the ECAT scanner electronics.

at 10 Hz. Past respiratory values are stored in an array and can be used for modifying respiratory state boundaries during a study. We have found, however, that the range of respiratory values drifts very little throughout a study, implying that a given respiratory A/D value should correspond to a specific rotated and translated heart position in the chest. Respiratory state boundaries are therefore usually fixed during the study.

The respiratory and cardiac states are used as indices in a 2D lookup table to select one of 16 possible histograms in scanner memory. TTL digital I/O lines on the NB-MIO-16 are used to encode the histogram selection as a 4-bit address. These four TTL outputs are connected to the ECAT front panel. At each cycle of the LabView sampling loop, the cardiac and respiratory states are determined and if necessary, the output state is changed via a digital signal over the TTL lines.

An example of a representative double gated study is seen in Figure 1. Here, the respiratory and cardiac cycles are both divided into five different phases. Cardiac phases are selected using the time boundaries with respect to each R-wave occurrence. For example, in one human study, cardiac time boundaries of 200, 300, 450 and 600 msec were used. The cardiac phases are grouped into three states: end diastole (0-200 msec and from 600 msec to the next R-wave), end systole (300-450 msec) and a transition state (200-300 msec, 450-600 msec). The five phases of the respiratory cycle are used directly to index five respiratory states in the 2D lookup table. Thus, the lookup table addresses 15 total locations in the histogram array. Note that the output state entries in the 2D lookup table need not be unique. In this example, the same output states were used to characterize cardiac phases 2,4 and 1,5.

Whether or not such an allocation of cardiac and respiratory states is the best scheme for reducing motion-induced blur is a difficult question to answer until we can better characterize the typical range of motion due to respiration and due to the cardiac cycle, and the amount of time spent in each position. Ideally, one would like to divide the acquisitions into as many states as necessary to stop both types of motion, however, both limited memory in the tomograph and limited statistics in the resulting reconstructed images prevents one from doing this. The division into a 2D array of histograms with 15 entries is therefore a compromise intended to reduce a significant portion of the motion blur.

In this prospective cardiac gating scheme, tomograph events are unbuffered. Therefore, the cardiac state must be a function of the time since the last R-wave, and possibly a function of past R-wave intervals as well. In the system's current configuration, no provision is made for bad beat rejection during cardiac gating. Data acquired during a mischaracterized cardiac state due to an irregular beat cycle will therefore degrade the reconstructed image. Fortunately, heart beat irregularities usually occur during end-diastole, when the heart is relatively stationary [9]. They probably do not add a great deal of blur to the resulting end-diastole images. In the future, we will utilize the flexibility of the LabView programming envi-

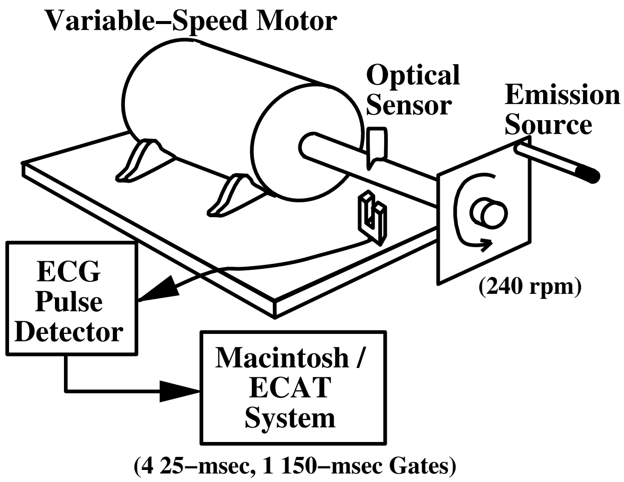

Figure 4: Schematic of rotating phantom. A ^{68}Ge source is attached to a rotating base and is imaged within the ECAT scanner. The optical sensor generates a waveform simulating a cardiac R-wave at the same position during each rotation.

ronment to investigate prospective bad beat rejection schemes.

Irregular breathing rates should not degrade respiratory gating at all in the current system configuration. Because the respiratory gate is based on the amplitude of the bellows signal, not the time since inhalation or exhalation, the duration of a breath makes no difference. Only the depth of respiration as detected by the respiratory bellows matters. We generally set up respiratory states so that they are evenly spaced for the range of bellows outputs during tidal breathing. Tomograph data acquired during spikes in the respiratory waveform, typically due to occasional deep breaths, are discarded. (This can be done, for example, by adding an extra respiratory state with a state boundary at the lower extreme of expected tidal respiratory values, and by setting the output value in this column of the 2D lookup table to a single unused histogram buffer). Note that it is still an open question whether the simple amplitude of the digitized bellows signal is the best measure of respiratory state. Lack of drift seen in the digitized respiratory waveform seems to indicate that it is a reasonable indicator. We are investigating validation techniques using MRI navigator echos [11] to answer this question.

IV. System Performance

Though the system is based on low-cost hardware not usually employed for real-time control, we found that with careful software design, adequate performance could be achieved. Maximum respiratory sampling rate for the system is approximately 250 Hz, or 4 msec per sample. Maximum sampling rate of the counter measuring time since the last R-wave is 1 KHz, or roughly 1 msec per sample. Changing states of the TTL output can be done in about the same time as an A/D conversion. All these I/O operations compete for the same CPU and register space on the Macintosh. Therefore, since the slowest operation is a respiratory A/D conversion, the system can ensure a correct output state with respect to the R-wave occurrence to within 4 msec (the time required for a respiratory sample).

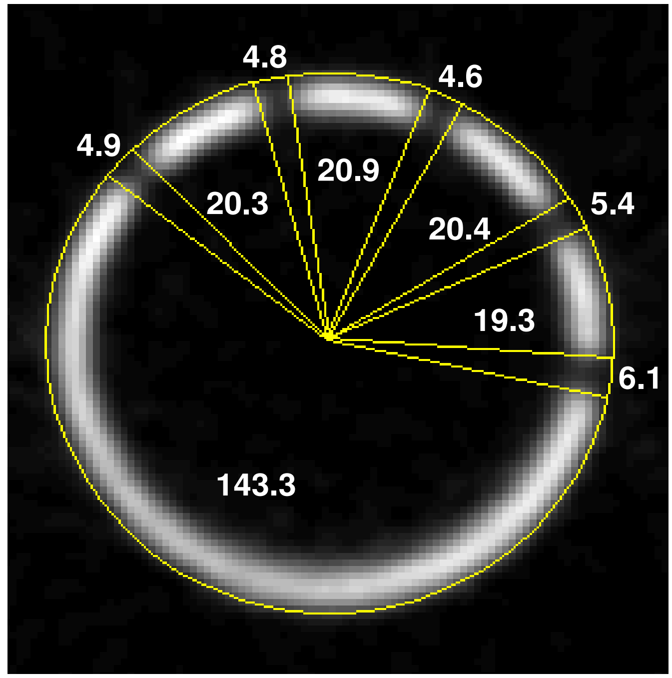

Figure 5: Composite reconstructed image of five states from the rotating phantom. Time in msec for imaging and switching periods is indicated on the image. Approximately 5 msec are required to switch states in the Macintosh/ECAT system.

Because of the infrequent respiratory samples, the state is usually switched with an uncertainty of only 1 or 2 msec (the time required for a cardiac counter/time sample).

Once the ECAT front panel receives a new TTL gating state, redirection of tomograph events from one histogram memory location to another is not instantaneous. Recognition of the state change and reconfiguration of scanner hardware result in lost and delayed events during state transitions. To measure these effects, as well as the effects of cardiac gate windowing uncertainty by the Macintosh front-end, a rotating phantom was built. The phantom rotates an emission source within the scanner's field of view and generates a pulse at the same point during each rotation (Figure 4). The pulse from the optical sensor was used in place of a ECG-derived pulse to drive cardiac gate changes in the data acquisition system. Data were acquired using a phantom revolution rate of 240 rpm, four 25 msec "cardiac" states, and a fifth gate that extended until the next sensor pulse. No output states were based on the inputs to the respiratory channel, but the A/D sampling of this channel was set to 10 samples/sec so that the overall system would see the same respiratory sampling delays as would occur during a typical patient study.

Reconstructed images from the resulting five sinogram sets show arcs corresponding to these portions of the circle which the emission source traveled during each gating period. Figure 5 displays a composite image of the five states and the time in milliseconds for approximate imaging and transition times. It is seen that approximately 5 msec of data were lost between each gate transition. Furthermore, the arc boundaries are well defined, indicating that the delay was repeatable, and hence did not temporally smear the data. This verifies the important fact that data during switching intervals is simply lost, and not redirected to an incorrect histogram. If the latter characteristic were true, unpredictable blurring would result.

Putting these delays in terms of switching times for a typical gated study on a patient, consider the 3x5 (i.e. three cardiac states, five respiratory states) protocol displayed in Figure 1 for a patient breathing at 15 breaths/minute with a heart rate of 80 bpm. Assuming every respiratory state is reached during every breath, a total of 8 output state changes would result from one inhalation/exhalation cycle; and 4 output state changes would result for each heartbeat. This corresponds to an average of 440 state changes per minute, or one state change every 136 msec (though due to the asynchronous nature of the cardiac and respiratory waveforms, states could change as frequently as the Labview loop sampling rate, every 1-4 msec.) We would expect roughly 2 seconds of data lost per minute in this example assuming a 5 msec switching time. Double gating in this case would result in a loss of less than 4% of the available data. In our studies thus far, we have seen roughly 2-3% of events lost due to switching based on the sum of resulting active time from a timed gated study.

V. Conclusions

The use of a Macintosh platform and LabView environment made this a low-cost system that was relatively easy to program. We have found that thus far, it has met our signal processing needs. On the other hand, our test results show that we are currently near the limit of the system's processing power. For example, tasks such as status displays to a LabView control panel on the Macintosh screen have to be minimized to avoid real-time acquisition delays. Also, in order to obtain the reported performance, low-level I/O subroutines were written rather than relying on high level LabView I/O modules.

Recent improvements in the system include replacement of the Macintosh IIci with a Macintosh Quadra 650. This faster computer increases the maximum respiratory sampling rate from 250 Hz to 550 Hz, and the maximum cardiac sampling rate from 1 KHz to over 3 KHz. These faster rates allow more computer processing between samples so that more sophisticated cardiac and respiratory state determination schemes may be investigated.

Future system improvements include the replacement of the ECAT ACS I control hardware with ACS II hardware. This upgrade will allow list-mode acquisitions, where tomograph events are tagged and stored individually to disk, rather than summed in just a few histograms. In this scheme, one of the 4-bit output states may be used to encode an R-wave occurrence, and the remaining 15 output states may then be utilized to encode respiratory position. The technique will remove the limit of dividing a doubly-gated acquisition into at most 16 histograms and will give the research advantage in that alternate gating strategies can be considered retrospectively, allowing trade-offs to be studied from a single acquisition.

VI. Acknowledgments

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