INTRODUCTION

Identification and segmentation of structures of interest are necessary steps in the computer-based analysis of medical images. Computer-aided diagnostic (CAD) systems utilize segmentation algorithms to isolate specific structures (represented by 2D or 3D regions in an image or set of images, respectively); conversely, to remove extraneous structures that may introduce errors in the computerized analysis. This step increases both the specificity and sensitivity of the CAD system and decreases computation time by focusing analysis on smaller regions representing the structures of interest. Segmentation of the lung parenchyma is often the first step when computerized analysis focuses on the thorax. High contrast, central positioning, relatively large size in comparison to other thoracic structures, and contiguous to other critical structures (e.g., heart) render the lungs useful as both a target for primary analysis and a reliable starting point for the analysis of other thoracic structures.

Segmentation of lung parenchyma in computed tomography (CT) scans is one of the most extensively researched areas in medical image processing. The low density of lung parenchyma compared with surrounding soft tissue translates into high contrast on CT, which in turn facilitates the use of several image processing techniques such as histogram thresholding, active contours, and seeded region growing. Conversely, multiple factors have limited the clinical utility of thoracic magnetic resonance imaging (MRI) and thus limited the need for lung segmentation in MR scans. Contrast and orientation of magnetic resonance scans are determined by the image acquisition protocol, and thus may require image processing methods specific to each pulse sequence and image orientation. The clinical utility of thoracic MRI is also limited by low resolution and long acquisition times that cause severe image artifacts. Recent improvements in the in-plane resolution, pulse sequences, acquisition time, and contrast media (e.g., hyperpolarized gas), however, have made MR a viable modality for thoracic imaging and have renewed interest in lung segmentation for thoracic MR applications (Eibel et al., 2003; Entwisle, 2004; Evans and Gleeson, 2004; Levin et al., 2001; Weber et al., 2004).
THORACIC MAGNETIC RESONANCE IMAGING AND ACQUISITION ARTIFACTS

A magnetic resonance scanner generates high-contrast soft-tissue images without subjecting a patient to ionizing radiation. First, a patient is placed in a strong magnetic field generated by a superconducting magnet. The nuclei of the hydrogen atoms that compose the tissue of the patient possess a small magnetic moment that causes the nuclei (essentially protons for hydrogen atoms) to align along and precess about the magnetic field. The patient is then subjected to a radio-frequency pulse that causes the hydrogen nuclei to temporarily rotate perpendicular to the axis of the magnetic field. In this alignment, the precessing hydrogen nuclei induce an electric current (signal) in a receiving antenna connected to the magnetic resonance scanner. This signal is then mathematically reconstructed into an image of the patient. The reconstruction maps the signal to a rectangular matrix of numbers, where the position in the matrix corresponds to the physical position of tissue in the patient, and the matrix value (also called gray-level value, brightness, or signal intensity) is proportional to the density of hydrogen nuclei at that specific position in the patient. In general, the greater the density of hydrogen nuclei in the patient, the brighter the gray-level value recorded in the image. The bone and lung parenchyma of a healthy patient produce almost no MR signal due to the low density of hydrogen nuclei. Conversely, thoracic soft tissue and diseases of the lung and pleura (e.g., tumor and effusion) contain a substantially higher density of hydrogen nuclei and thus exhibit high signal intensity. Thus, abnormalities of the lungs in MR images will appear as bright, high-signal tissue on the surrounding dark, low-signal lung parenchyma background. The introduction of intravenous gadolinium contrast further increases the image contrast by increasing the signal of diseased tissue without impacting the signal of normal lung parenchyma.

The promising soft-tissue contrast properties of thoracic MR imaging are mitigated by several acquisition artifacts that reduce image quality and may introduce errors into CAD analysis. Ghosting is the periodic repetition of a structure along the phase-encoding dimension of an image due to motion during image acquisition. In thoracic imaging, several structures (e.g., lungs and heart) exhibit repetitive motion during image acquisition that may result in ghosting (Hashemi et al., 2004; Liang and Lauterbur, 2000). Cardiac motion artifacts occur due to the beating of the heart and the pulsation of blood through the major vessels. In transverse MR sections, this artifact manifests as a large column of noise (ghosting) that extends along the anteroposterior dimension of the image (Figure 14.1). This column often masks the underlying lung parenchyma and may mimic disease due to its high signal intensity. Injection of a contrast agent such as gadolinium further increases the signal intensities of the heart and thus exacerbates cardiac motion artifact. Pulmonary motion artifacts are related to respiratory motion and manifest in two ways. First, the expansion and contraction of the thorax may cause ghost contours of the chest wall to form both inside and outside the patient thorax.