Automated analysis of 3D echocardiography

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Introduction 1

1.1 | Motivation

cardiovascular disease

Cardiovascular disease has been the number one cause of death in the world for the last decades and is projected to remain the leading cause of death [WHO 2007]. Cardiovascular disease encompasses disorders of the heart and blood vessels, either originated at birth (congenital heart disease) or developed during life. Among the latter are atherosclerosis, such as coronary artery disease (possibly causing a heart attack) and cerebrovascular disease (causing stroke), arrhythmias, hypertension, heart failure and many other diseases. Major (modifiable) risk factors for cardiovascular disease include unhealthy diet, physical inactivity and tobacco use.

left ventricle of the heart

In this thesis we will focus on the assessment of global functioning of the left ventricle of the heart. This ventricle is responsible for pumping the blood, coming from the lungs, where it is saturated with oxygen, through the whole body. Measurement of the left ventricular volume and function is therefore very important in clinical decision-making, assessment of therapeutic effects and determination of prognosis.

Malfunctioning of the left ventricle may be caused by coronary artery disease, hypertension or arrhythmias. Ischemia, which eventually results in heart failure may cause a wide variety of symptoms.Since in mild cases of heart failure symptoms may be faint and a universally agreed definition is lacking, the disease is often undiagnosed. This may have severe consequences, including even death.

3-dimensional echocardiography

While prevention aiming at reduction of the main modifiable risk factors can reduce the number of deaths caused by cardiovascular disease, also a wider availability of diagnostic techniques might improve the treatment of cardiovascular disease. Since the beginning of this century, 3-dimensional echocardiography (3DE) has become available and is getting more widespread across medical centers. 3DE offers a non-invasive, relatively cheap and therefore possibly widely available way to visualize the left ventricle in 3D (fig. 1.1) and to analyze its function. However, manual analysis of these images for quantitative assessment of functional parameters is cumbersome. Therefore, automation of the assessment of left ventricular function is an important step in improving the diagnosis and treatment of cardiovascular disease and reducing its costs.

automating the analysis

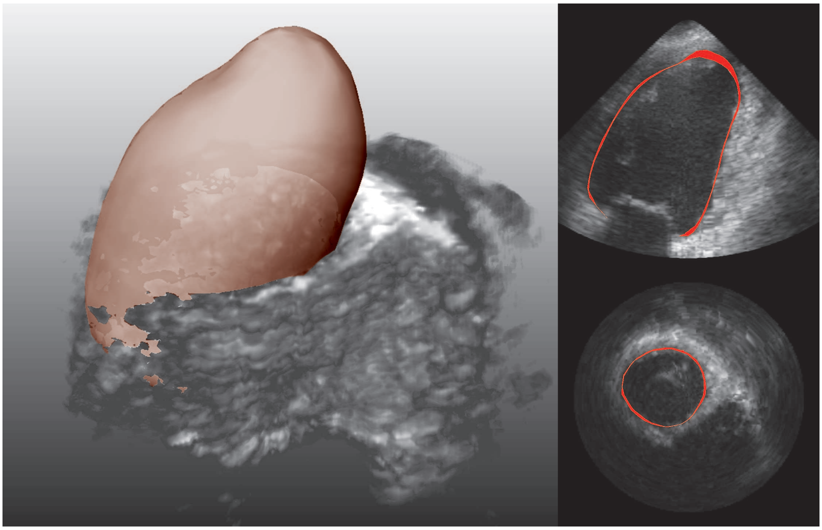

Figure 1.1: A 3D ultrasound image of the left ventricle with the endocardial surface in blue. *left*) A 3D rendering of the image with the endocardial surface semi-transparant in red. *top right*) A 2-chamber view of the same data set, with the delineation of the endocardial border. *bottom right*) A short-axis view of the same data set

3D echocardiography | 1.2

Imaging of the heart poses many challenges on imaging modalities. To assess the functional parameters of the heart, the geometry and dynamics of the heart should be imaged in great detail. Therefore, ideally, the full cardiac cycle is imaged in real-time, distinguishing different types of tissue with high spatial and temporal resolution, and with minimal discomfort for the patient, at low costs.

In the past decade, echocardiography (ultrasound imaging of the heart) has been conquering many technological challenges to achieve this goal. It has been developed into a very competitive imaging technique with its own strengths and limitations. In this section we will discuss these characteristics from a technical and clinical point of view.

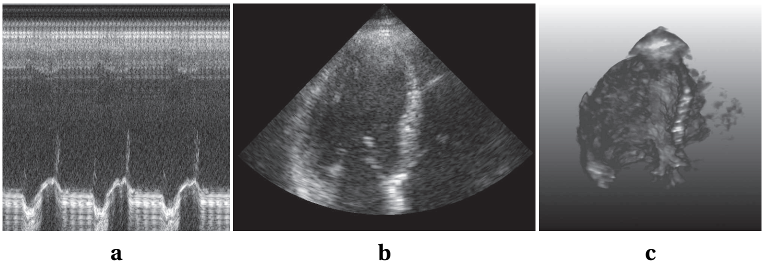

Figure 1.2: Different acquisition modes for echocardiography. *a*) An M-mode image, showing one acquisition line over time (horizontal axis) *b*) A B-mode image, showing a cross-section of the left ventricle. *c*) A 3D rendering of the left ventricle, acquired by mechanically rotating a phased array transducer around its central image axis

1.2.1 | Ultrasound imaging

principle

Imaging using ultrasound is done by transmitting beams of high frequency sound and recording the resulting echos. Echos come from transitions between materials of different densities, for example air and bone, or blood and tissue. The larger the density difference, the higher the intensity of the echo. Given the speed of sound in the imaged object and the recorded time between sending the sound pulse and receiving the echo we can locate the reflector or scatterer. When imaging tissue, the echos are a result of scattering of the ultrasound beam due to the inhomogeneous nature of the tissue, generating speckle.

Ultrasound is transmitted by applying an electrical field on piezoelectric material (the transducer element), making it vibrate at high frequency and transmitting ultrasound. Receiving echos is basically the same process in reverse. During traversal of the ultrasound beam through the medium, multiple echos from various depths can be recorded, generating an image line (A-mode image). Imaging this line over time gives an M-mode (motion) image (fig. 1.2a).

making an image

A 2-dimensional (2D, B-mode) image is built up from image lines that are recorded by sending and receiving focused sound beams under different angles. This can be done by sweeping the ultrasound beam mechanically (fig. 1.2b).Transmitting a beam under a certain angle can also be done by using multiple sound sources (the transducer elements) and activating them with a time delay between neighboring elements such that a sound wave in the desired direction is created that converges at a certain depth (fig. 1.3). Such an array of transducer elements used for 2D imaging is called a phased-array transducer. Phased-array transducers are most commonly used in 2D echocardiography.

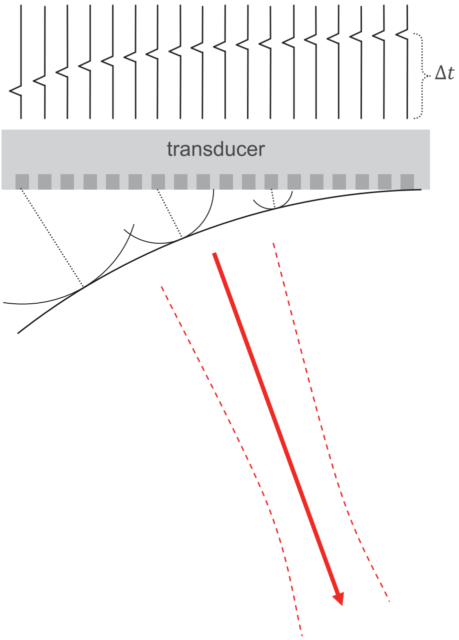

Figure 1.3: Electronic beam steering using a phased array transducer. The transducer elements (dark gray) are activated with such time delays (Δt) that a wave front (solid redline) under the desired angle is created

major parameters

A few of the most important parameters that determine the quality and resolution of the recorded images are discussed in this paragraph. The material and thickness of the transducer elements determine the resonance frequency and bandwidth of the transducer. The higher the frequency, the smaller the penetration depth, but the higher the axial imaging resolution. The speed of sound is assumed to be constant in human tissue and limits the number of beams that can be sent and received sequentially per time unit, restricting the number of frames that can be imaged per second (the frame rate). The width of the array of transducer elements influences the width of the focussed beam and is therefore a limiting factor in the lateral resolution of the image. In adult 2D echocardiography, typical resolutions are 0.3 mm axially and about 1° laterally.

harmonic imaging

Since a decade, tissue harmonic imaging [Spencer et al. 1998; Tranquart et al. 1999] has been widely adopted in medical ultrasound imaging. Due to nonlinear propagation of the ultrasound wave through the tissue, higher frequencies, harmonic modes of the transmitted signal, are generated. To exploit this phenomenon, transducers need to be designed that are sensitive to these higher frequencies. Previously, transducers were optimized to transmit and receive in the same frequency. Because of the higher frequency, a higher resolution can be achieved in the image [Ward et al. 1997]. Furthermore, harmonic imaging shows some advantages over fundamental imaging which result in clearer images and reduced near-field clutter[Duck 2002; Thomas and Rubin 1998].

1.2.2 | Developments in 3D echocardiography

Conventional 2D echocardiography (B-mode imaging of the heart) allows visualization of a slice of the heart over time and is widely used for assessment of cardiac function. 2D echocardiography allows measurement of left ventricular volume and derived parameters such as ejection fraction, stroke volume and cardiac output. However, assumptions about the left ventricular geometry and the position of the imaged planes in 3D, need to be made.

Therefore, ever since the existence of 2D echocardiography, people have been searching for an extension to 3D, to overcome the limitations of 2D echocardiography [Bruining et al. 2000].

1.2.2.1 | Freehand 3D imaging

First attempts towards 3D echocardiography were made by freehand scanning using a conventional 2D transducer that was registered in 3D either acoustically (so-called spark gap location [Moritz and Shreve 1976]), using a mechanical arm [Dekker et al. 1974] or using electromagnetic spatial locators [Barratt et al. 2001; Raab et al. 1979]. 3D image reconstruction is performed offline by dedicated software. Only the electromagnetic tracking systems eventually made it into the clinic, because of practical limitations of the acoustic and mechanic positioning systems. But still, also electromagnetic freehand 3DE has its limitations. The positioning accuracy is limited and the acquisition is time-consuming and cumbersome. It suffers from motion artifacts as a result of patient movement and breathing. In spite of these restrictions, freehand 3DE has been used till recently [Mannaerts et al. 2003; Varandas et al. 2004] because of its cost-effectiveness.

1.2.2.2 | Mechanical 3D imaging

To shorten acquisition times and improve on irregular coverage of the 3D space by freehand acquisitions, the acquisitions have been automated in several ways.

The first approach is a linear scan of the target space resulting in parallel 2D

linear

images constituting the 3D volume (fig. 1.4a). Pandian et al. explored, among

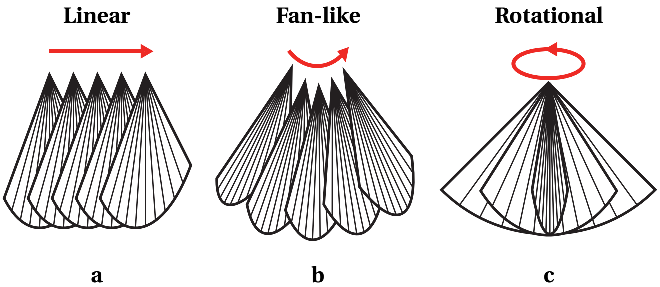

Figure 1.4: Mechanical 3D scanning modes. *a*) Sweep mode. *b*) Fan-like mode. *c*) Rotational mode

other configurations, the possibilities of computer-controlled serial 2D cardiac tomographic images extensively [Schwartz et al. 1994].

Secondly, mechanical fan-like sweeping of the phased-array has been proposed (fig. 1.4b). In this way, a pyramidal volume can be scanned by moving the transducer in a fan-like arc at prescribed angles [Delabays et al. 1995]. In contrast to the linear scan approach, fan-like movement of the transducer is more suitable for transthoracic echocardiography, because of the limited echo window. *fan-like*

Thirdly, stepwise rotational scanning of the volume has been applied (fig. 1.4c), where the phased array is rotated around its central image axis, such that co-axial images are acquired resulting in a conical 3D data set [Pandian et al. 1994]. The number of co-axial planes can be varied to prioritize between the speed of the acquisition and the accuracy of the imaged volume, and the derived clinical parameters [Nosir et al. 1996; Papavassiliou et al. 1998]. *stepwise rotating*

Finally, pseudo real-time approaches have been presented where a phased array is continuously rotated internally around its central image axis (fig. 1.4c) [Belohlavek et al. 2001; Canals et al. 1999; Djoa et al. 2000]. For both approaches by Belohlavek et al. [2001] and Canals et al. [1999] the rotation direction is periodically alternated to prevent the cables from getting damaged. The acquisition durations are limited. This provides enough data for 3D LV volume quantification, but restricts the volume reconstructions to low frame rates. *continuous rotating*

The design by Djoa et al. [2000] has been extended to harmonic imaging by Voormolen et al. [2006]. A prototype of this transducer has been used in all studies in this thesis. It features a phased array that is continuously rotated at high speed in one direction, employing a slipring construction (fig. 1.5). This fast rotating ultrasound (FRU) transducer (see section 4.1.1) allows long acquisitions up to 10 seconds, which are used for pseudo real-time 3D volumetric reconstructions (fig. 1.2c). Temporal resolutions up to 25 phases per cardiac cycle are achieved, independent of the patient's heart rate.

Figure 1.5: The fast rotating ultrasound (FRU) transducer

1.2.2.3 | Real-time 3D imaging

Truly real-time 3D echocardiography has been realized originally at Duke University. Von Ramm et al. were the first to build a matrix transducer for real-time 3D imaging [Smith et al. 1991; von Ramm et al. 1991]. Subsequent developments by this group led to the first commercially available 3D phased-array system at the end of the 90's (Volumetrics Model 1, Volumetrics Medical Imaging, Durham, NC).

current 3DE systems

Second generation matrix transducers were introduced by Philips Medical Systems (Best, the Netherlands) and later by General Electric (Milwaukee, Wisconsin, USA). The Philips Sonos 7500 scanner with a X4 xMatrix transducer is capable of live imaging a narrow volume of 25° × 90° at a frame rate of 25 Hz. Full volume imaging is achieved by stitching four narrow image sectors, acquired from seven consecutive beats, together into one volume. It has been succeeded by their Sonos iE33 system with its X3-1 transducer, which shortens full volume imaging to only four cardiac cycles. The same approach is followed by General Electric with their Vivid 7 scanner and its 3V transducer. Recently, also Toshiba and Siemens announced their 3DE systems, of which the latter claims to be able to do real-time imaging of a 90° × 90° volume at 20 Hz, eliminating the need of any ECG gating in the 3D acquisition.

matrix advantages

A major advantage of the real-time scanners is their ability to show live 3D renderings, while acquiring data. Pseudo real-time solutions using mechanically rotated phased-array transducers (such as the FRU transducer) rely on off-line analysis of the acquired data to achieve a 3D rendering.

Advantages of the FRU transducer over matrix transducers are its better image quality in the 2D image frames and its cost-effectiveness. Furthermore, the FRU transducer allows reliable quantitative analysis based on single-beat data, featuring 6 to 8 2D images per cardiac phase (if using 16-20 phases per cycle) [Voormolen et al. 2007].

FRU advantages

Clinical application | 1.2.3

3D echocardiography has some clear advantages over 2DE in the clinical environment.

At first, the standard 2DE apical views (2-chamber, 4-chamber, long-axis view) can be acquired at once, reducing the acquisition time. Also, 3DE does not suffer from foreshortening because anatomical plane selection can be done off-line, resulting in true standard views. Furthermore, any plane can be visualized off-line.

Secondly, the full geometry of the left ventricle can be imaged. This eliminates the need of making assumptions about the LV geometry in quantitative analyses. This allows more accurate estimation of clinically important parameters such as full cycle LV volume and its derived parameters (ejection fraction, stroke volume, cardiac output, etc.) [Jenkins et al. 2004]. Also, better insight in the LV and the valve geometry is given through 3D renderings of the left ventricle, including better visualization of the wall motion and possible abnormalities. This makes 3DE also a very promising successor for routine stress echocardiography, since regional wall motion abnormalities can be located much more accurately.

Stress echocardiography | 1.2.3.1

Another application of 3D echocardiography involves stress echocardiography. In stress echocardiography patients are examined at different stages of physical or pharmacologically induced stress to visualize regional wall motion abnormalities as a result of myocardial ischemia. 2D stress echocardiography (2DSE) has become a well established tool for identification of patients with coronary artery disease [Armstrong and Zoghbi 2005; Geleijnse et al. 1997]. 3D stress echocardiography (3DSE) has shown to improve on several limitations of 2DSE, such as better anatomical plane selection for comparison of identical wall segments in the different stress stages. Current limitations of 3DSE however, include serious drop outs in the LV lateral wall from rib shadowing, limited temporal resolution and stitching artifacts as a results of volume stitching. All these limitations are expected to be handled by technical developments in 3DE, resulting in smaller transducer footprints,

3D stress echocardiography

larger bandwidth transducers suitable for harmonic imaging and higher temporal resolution as a result of parallel beam forming.

1.2.3.2 | Limitations

limitations

Current limitations of 3DE are its slightly compromised image quality if compared to 2DE. 3DE spatial image resolution is lower and sensitivity of matrix transducers is still inferior to 2DE. This results in typical image artifacts, such as serious drop outs in the lateral wall region. The temporal resolution of 3DE is also much lower than that of 2DE, which limits the use to patients with relatively low heart rate if reliable estimation of the volume-time curve is needed. Furthermore, high costs are associated with 3DE, which makes 3DE much less commonly available. Also, 3D imaging of the heart has shown to require adequate training of the sonographer.

Most of the limitations mentioned above are expected to be tackled soon, as we currently see rapid developments in transducer design that allow real-time imaging of larger volumes and higher frame rates, with better image quality and resolution. It is to be expected that eventually 3DE will replace 2DE in clinical routine examinations.

1.2.4 | 3DE vs. other modalities

Because of its cost-effectiveness, echocardiography is an attractive imaging modality. It is usually widely available and ultrasound devices can be made portable, allowing bed-side imaging. Furthermore, ultrasound imaging is non-invasive and does not employ ionizing radiation. No adverse biological effect has been reported so far, provided that guidelines for use of diagnostic ultrasound are respected [Barnett et al. 2000].

In diagnosing cardiovascular disease, various other imaging modalities are available, each with its specific strengths and limitations. We will discuss the most common other modalities in image guided diagnosis of cardiovascular disease.

1.2.4.1 | Cardiac magnetic resonance imaging

golden standard

Magnetic resonance imaging (MRI) is an important imaging technique that allows non-invasive 3D imaging of the human body. MRI is especially suitable for imaging of soft tissues. The in-plane spatial resolution of current 1.5T scanners is approximately 1.5 × 1.5 mm, which is comparable to 3D echocardiography. The high temporal resolution of more than 30 frames per second, when using steady-state free precession sequences, combined with the high contrast resolution, makes MRI very suitable for imaging of the heart. Therefore, MRI is accepted as the golden

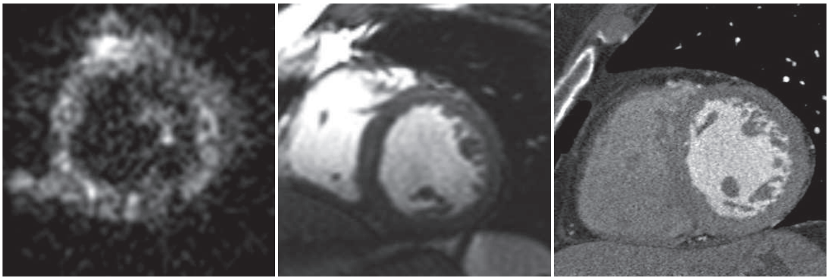

Figure 1.6: Example images from different patients showing several modalities for imaging the left ventricle. *a*) A short axis slice from a 3D ultrasound image *b*) A short axis MR image *c*) A short axis CT image.

standard for assessment of left ventricular function (fig. 1.6b).

Just as for ultrasound, various different pulse sequences can be used, to target the imaging protocol to specific tissue types or physiological processes. This makes MRI suitable for one-stop-shop approaches, in which various clinically important parameters can be assessed in a single, although possibly time-consuming, scan session. Such a one-stop-shop session might include assessment of LV and RV (regional) functional parameters, myocardial perfusion imaging and late enhancement MRI for localization of infarcted regions and assessment of myocardial viability. *one-stop-shop*

Despite its favorable image quality and its harmless nature, MRI is not used as the main imaging modality in cardiology because of the time needed for imaging, the high costs associated with the systems and therefore, their limited availability. Furthermore, the high magnetic fields and powerful radio frequency pulses prevent its use on patients who have metal implants and cardiac pacemakers. *limitations*

If compared to 3D echocardiography, several issues have to be kept in mind. The high resolution in the 2D MRI (typically short-axis) images are compromised with a poor through-plane resolution of up to 10 mm. In assessment of left ventricle function this is especially an issue when it comes to defining the base of the left ventricle. This, together with the partial volume effect [Lorenz et al. 1999], different appearance of papillary muscles and trabeculae and different (semi-automated) analysis methods hamper the direct comparison of assessments by MRI and 3DE [Voormolen and Danilouchkine 2007]. *MRI vs. 3DE*

1.2.4.2 | Computed tomography

Computed tomography (CT) is an imaging technique based on X-ray imaging. A large number of 2D images is taken around a fixed rotation axis to reconstruct a 3D volumetric image. Current multi-slice CT (MSCT) scanners employ up to hundreds of detector rings to acquire the 3D volume faster and allow a higher temporal and spatial resolution. The high temporal resolution opens doors for functional assessment of the heart (fig. 1.6c), and for coronary angiography and perfusion studies.

The main drawback of MSCT, in comparison to MRI and 3DE, is the radiation exposure that is associated with the acquisitions. This will limit the use of MSCT for (global) functional assessment of the left ventricle. Coronary angiography and calcium scoring might be of more interest, although recent studies show that if it comes to ruling out coronary artery disease (CAD), the high negative predictive value of MSCT is compromised with a moderate positive predictive value [Schuijf and Bax 2008].

1.2.4.3 | Nuclear imaging

SPECT Single photon emission computed tomography (SPECT) is a 3D technique that images the distribution of an, intravenously injected, radiopharmaceutical in the body. It can be used to assess myocardial perfusion during different stages of physical or pharmacologically induced stress [Bax et al. 2000; Corbett and Ficaro 1999]. The resolution of this technique is limited and morphological information about the left ventricle is poor. But differences in myocardial perfusion between the stress and rest stage reveal valuable diagnostic information about infarcted regions [Corbett and Ficaro 1999]. Integration with CT allows registration of the perfusion data to the morphological CT images.

PET Positron emission tomography (PET), is a similar technique that also images the distribution of a radiopharmaceutical that indicates tissue metabolic activity. It can be used to detect coronary artery disease with high sensitivity and specificity [Williams 1994]. PET scanners can, like SPECT, be integrated with MR or CT. However, PET has a limited role in routine diagnose of myocardial defects, because of the high costs associated with the production of the necessary radionuclides.

1.3 | Digital image analysis

Images are everywhere. Recently, digital cameras have become so widely available that they did not only replace traditional analog cameras but also got integrated in mobile phones, PDAs, laptops and other handheld devices. In medicine, a similar progression has been going on. Traditional analog X-ray systems are being replaced by their digital counterparts and echocardiography tapes are being replaced by CD's, DVD's and hard disks. Other complicated digital imaging techniques have become available thanks to developments in computer science (MRI, CT). Nowadays lots of different imaging modalities are available to assist in diagnosis. So many images are acquired, that automation in acquiring, reconstructing, enhancing and analyzing them has become essential.

1.3.1 General purpose of automated image analysis

Of course, images are acquired primarily for visual inspection and to get insight into the anatomy and physiology of the organ of interest. But there is a growing demand towards techniques that can automatically analyze all these images and derive as much quantitative information as possible from them, such that image guided diagnosis and treatment is brought to a higher level and made more efficient and reproducible.

Automated analysis techniques in general aim at decreasing interobserver variability by ruling out random variability and judgment differences of human experts, making the analysis more reproducible and comparable among institutions. This may lead to a high degree of standardization, which eases the design of protocols and decision making. A high reproducibility is important if the progression of a certain disease is monitored over time. Also, the speed of the analysis can be improved and thereby the labour intensiveness and the costs of the diagnosis or treatment is reduced.

decreasing interobserver variability

reduce costs

Image analysis improves diagnosis and treatment by quantification of observations, either with or without human intervention in the analysis. Quantification gives more insight in the decision process and can ultimately lead to automated diagnosis, assisting the physician.

quantification

Since manual analysis of 3D echocardiography is cumbersome and very labour intensive, we aim at automating the analysis of 3D echocardiography. We try to automatically quantify the functioning of the left ventricle, to reduce interobserver variability and improve the reproducibility of the quantification results.

This work encompasses the reconstruction of the 3D image over time, from a sequence of 2D images for proper visual inspection of the 3D (plus time) data. Also, this image reconstruction acts as a preprocessing step to allow generalized algorithms for analysis of 3D (plus time) images. For quantitative functional analysis we aim at automated tracking of feature points and structures, to visualize and quantify change in position, size or shape, and orientation of these elements over

3D image reconstruction

segmentation time. This functional analysis also encompasses image segmentation, the automated detection of structures in the 3D image sequences. In these tracking and segmentation algorithms knowledge about image acquisition, the specific patient or the patient population and the targeted structure (or organ) is used to optimize its performance. In the next section we will discuss several important issues in automating these procedures in echocardiography.

1.3.2 | Automated analysis of echocardiography

Unlike other tomographic modalities such as CT and MRI, ultrasound images are hard to interpret, since there is no simple physical relation between the observed image intensity and the imaged medium. Interpreting 3DE is therefore not only a challenge for the untrained human eye, but even more for automated image processing techniques. In the automated analysis of 3DE we have to deal with several ultrasound specific image characteristics.

1.3.2.1 | Image characteristics

Ultrasound image gray values are a result of a summation of sound reflections and scattering, resulting in a combination of interference patterns, called speckle patterns. *speckle* These patterns give a granular appearance to the image. Differences in imaged media or tissues are observed through differences in these speckle patterns and their intensities. Therefore, transitions between different types of tissue need not render a clear edge in the image, but might show only subtle differences. This granular appearance of the image might challenge the interpretation of the image, but can be of great value when imaging translations and deformations and make ultrasound very suitable for tracking approaches.

The object appearance is also position dependent in ultrasound imaging. The *position dependency* signal depends on the depth and the objects in the line of sight. Acoustically dense structures might drop a shadow on regions further away from the transducer. Attenuation can be compensated manually using time gain compensation (TGC) while acquiring, but lower signal-to-noise ratios in distant image regions can, of course, not be compensated. The angle-of-incidence of the ultrasound beam influences the reflection and scattering and because of the fanlike acquisition of subsequent beams, echocardiographic images are also highly anisotropic.

Several other image artifacts can be caused by side and grating lobes, reverberations, aberration and noise. Some of these artifacts might be reduced by using harmonic imaging [Duck 2002].

Considerations for automated analysis | 1.3.2.2

For as long as echocardiographic images have been made, also attempts to automatic analysis strategies have been reported. An overview of quantitative methods in 2D echocardiography has been given in Bosch [2007]. Noble and Boukerroui [2006] published a general review of ultrasound image segmentation, also for non-cardiac applications. We will shortly discuss the main considerations when developing an automated analysis approach, relevant to the subject of this thesis.

Ultrasound image appearance is characterized by its granular appearance from speckles and its artifacts as described in the previous section. A great advantage of echocardiographic imaging, at least in 2D, is its high frame rate. These three aspects (speckle, temporal information, typical artifacts) should be considered when designing an analysis technique, whenever possible.

speckle as a feature

Image speckle can be used as a local image feature. On a small scale speckle serves as a distinct image feature that can be exploited by tracking approaches, as long as object movement is small, relative to the speckle size. This is often employed in 2D echocardiography [Behar et al. 2004; DeCara et al. 2005] It should be noted however, that speckle patterns depend on the imaging system and that they can change considerably as a result of deformation of the tissue or change in orientation with respect to the transducer. Despite these limitations, texture characterization has been successful in various ultrasound applications [Christodoulou et al. 2003; Sivaramakrishna et al. 2002; Yoshida et al. 2003]. On a larger scale speckle might be an undesirable feature, resulting in a non-Gaussian gray value distribution. Various models have been presented that model the gray value distribution in ultrasound images, which can be incorporated into the detection method [Mignotte and Meunier 2001]. Alternatively, a preprocessing step is often applied, which removes speckle from the image and possibly also aims at transforming the gray value distribution into a Gaussian distribution, such that more general image processing approaches that rely on this property can be applied [Tauber et al. 2004; Xiao et al. 2004; Yu and Acton 2002].

speckle suppression

time domain

An important source of information comes from the temporal domain. In this domain we can identify static image artifacts, for example as a result of rib shadowing or near-field clutter, and remove noise, such as in the far field. Apart from identifying image artifacts, the temporal information provides most of the functional information we want to extract from an image sequence. The temporal domain can be exploited as multiple observations of a static scene. In this way the object's dynamics are just observed without enforcing any constraints on the dynamics. More robust detection solutions model the object's dynamics to constrain the motion and deformation to expected behavior, as has been elegantly employed in Friedland and Adam [1989] and Comaniciu et al. [2004].

Because of the typical artifacts that are present in ultrasound images, methods that are solely based on local image features are prone to fail. Image artifacts should be actively detected based on regional spatiotemporal image information, such that a reliability measure can be integrated into the detection [Zhou et al. 2005]. Alternatively, higher level knowledge about the object to be tracked or segmented can be incorporated into the method. This information can be provided by the user, but is ultimately integrated into a model. Various knowledge or model based techniques can be applied to deal with these typical artifacts. Knowledge can be integrated by using some simplified mathematical model to represent the shape of the object, for example based on geometrical assumptions. But also the expected image intensities can be modeled as such, as well as temporal behavior of the object. Various methods integrate prior information about shape and texture [Montagnat and Delingette 2000; Xie et al. 2005]. All of these object properties can also be learned from a training population, for example using a neural network approach as in Binder et al. [1999]. Another concept that is capable of modeling such properties are statistical models, for example active appearance models. These models have been successfully applied to detection of endocardial contours in 2D echocardiographic sequences [Bosch et al. 2002].

image artifacts

model based detection

In chapter 6 we will explore the application of active appearance models for automated segmentation of the left ventricle in 3D echocardiography. In this chapter we also review most of the approaches that have been presented for automated segmentation in 3D echocardiography.

1.4 | Outline of this thesis

Automated estimation of left ventricular volume has been the subject of research for many years. The recent developments in real-time 3D echocardiography have made the assessment of full cycle left ventricular 3D images feasible as a quick, non-invasive, relatively cheap and therefore potentially widely available technique. Manual analysis of the 3D time series of these data sets, however, is cumbersome and subjective, and therefore causes relatively high inter- and intraobserver variabilities in quantifying global left ventricular function. This limits the application in large, inter-institution clinical trials and hampers the value in diagnosis.

We have been challenged by the possibility of assessing global left ventricular function by real-time 3D echocardiography and by the success of previous model-based automated detection attempts to estimate the desired parameters from 2D echocardiography. This has been an inspiration for further improvement of automated assessment of these important clinical parameters using the opportunities that are provided by the recent innovations in ultrasound imaging.

The main goal of our work therefore is the automation of left ventricular volume quantification using model-based segmentation in 3D echocardiography. This will improve the ease of use of real-time 3D echocardiography for assessment of important clinical parameters for diagnosis of left ventricular function. It will save costly time in analysis of the increasing number of clinical assessments and improve the availability of such parameters in daily clinical practice, with high accuracy and precision, thereby allowing better diagnoses. In this work we have investigated the use of the fast rotating ultrasound (FRU) transducer for real-time 3D echocardiography, which combines advantages of conventional 2D echocardiography with the hugely improved insight given by 3D echocardiography, keeping the general applicability of the developed image analysis techniques in mind.

the main goal

As a first step to a supervised, fully automatic technique we have worked on a semi-automatic solution for left ventricular volume quantification, which detects full-cycle volumes using limited user interaction based on 2D endocardial border detection in a four-dimensional framework. The interactive nature of this technique allows rapid segmentation of the left ventricle with high accuracy. This is a requirement for the development of a supervised fully automatic technique. The challenges in the extension of previous work to application in higher dimensions and the evaluation of this method have been described in chapter 2.

Chapter 3 studies an important element in full cycle left ventricular volume measurement: tracking the position of the mitral annulus. A substantial time reduction in these full cycle analyses could be achieved by automatic tracking of this quickly displacing anatomical structure. We have studied this problem in 2D echocardiography, with possible application in 3D echocardiography. We present a tracker for 2D structures over time, assuring a time-continuous solution for the mitral annular movement.

For the endocardial detection using native 3D or 4D imaging techniques we have studied the interpolation of the sequence of 2D images acquired in 3D within several consecutive cardiac cycles using the FRU transducer. This work, which deals with multi-beat fusion and the sparse, irregular distribution of the data, is described in chapter 4. An improved method is presented for interpolation of these numerous 2D images from consecutive cardiac cycles into one high resolution 4D cycle.

These high resolution reconstructions of the left ventricle allow native 3D or 4D model-based segmentation approaches for the detection of left ventricular volume. A common problem in the use of model-based segmentation techniques is the initialization of such models in a new data set. We have investigated the rarely studied subject of detection of the main orientation of the left ventricle in 3D acquisitions for initialization purposes. Chapter 5 discusses a technique for automated detection of the left ventricular long axis and the mitral valve plane. Knowledge about the position of these structures may improve model-based segmentation techniques significantly, since the performance of these techniques often depends on the quality of its initialization.

Chapter 6 presents a fully automatic segmentation technique for the estimation of left ventricular volume based on active appearance models. In this chapter we discuss the adaptation of these models to 3D echocardiography and explore the applicability of active appearance models with different matching approaches.

Finally, we conclude this thesis in chapter 7 and discuss the presented work with recommendations for future research in this direction.