Abdominal Wall Vibration Analysis for Evaluation of Biomechanical Properties and Physiological Diagnosis of Internal Abdomen

Name: Teunis Schuurman

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Coach: prof. dr. ir. D.J. Rixen
Professor: prof. dr. ir. D.J. Rixen
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Abdominal Wall Vibration Analysis for Evaluation of Biomechanical Properties and Physiological Diagnosis of Internal Abdomen

Design of a Measurement Setup and Evaluation of Data Analysis Methods

May 25, 2011

Supervisors:
Prof. dr. ir. D.J. Rixen
Dr. J.W. Hinnen
Dr. ir. C.A. Swenne

Author:
T. Schuurman
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<td>T. Schuurman</td>
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<tr>
<td>Email address</td>
<td><a href="mailto:teunis.schuurman@gmail.com">teunis.schuurman@gmail.com</a></td>
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<td>Student no.</td>
<td>WB1175572</td>
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<td>Precision and Microsystems Engineering</td>
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<td>Faculty</td>
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<tr>
<td>University</td>
<td>Delft University of Technology</td>
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<tr>
<td>Examination board</td>
<td>Prof. dr. ir. D.J. Rixen (TU Delft Dep. of Precision &amp; Microsystem Engineering)</td>
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<td></td>
<td>Ir. S.N. Voormeeren (TU Delft Dep. of Precision &amp; Microsystem Engineering)</td>
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<td>Prof. dr. J. Dankelman (TU Delft Dep. of Biomechanical Engineering)</td>
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<td></td>
<td>Dr. J-W Hinnen (LUMC Surgery)</td>
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<td>Dr. ir. C. Swenne (LUMC Dep. of Cardiology)</td>
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Preface

This thesis reflects the work carried out as a Master thesis assignment for the master Mechanical Engineering, Precision and Microsystems Engineering, at the Delft University of Technology.

The assignment for this thesis originated from, what Prof. Daniel Rixen would call, a ‘wild idea’, based on his mental leap that it should be possible to evaluate physiological properties within the human body by applying engineering dynamic analysis techniques on its interfaces. With the emergence of a student who was daring (or just stupid enough) to endeavor on this quest, a Thesis project was started in order to assess the usability of this approach.

Along the course of an earlier research project, a very pleasant cooperation was established between Prof. Rixen, Dr. Jan-Willem Hinnen and the present writer. Dr. Hinnen, specialized in the field of Abdominal Aortic Aneurysms (AAA), Endovascular Aneurysm Repair and the effect of intra-sac pressurization on aneurysm behavior, showed interest in the idea and was curious to see if the proposed technique could prove to be useful in detection and surveillance of abdominal aortic aneurysms. As such, he was more than willing to share his experience during the practical parts of this research and provided the opportunity to perform measurements on real AAA patients.

Dr. ir. Cees Swenne proved to be invaluable during the research. Not only was he able to provide an ECG monitor and later on suggested the use of a finometer for use in the definition of a frequency response function. He was also more than willing to spend time on instructions, to assist in the realization of proper measurements and was genuinely interested in the results.

I would like to thank Dennis de Klerk and Paul van der Valk for their cooperation, critical remarks and support along the way. Also the cooperation of Polytech, and especially the personal efforts of Frank Sven, who where able to make a license for the time module temporarily available for our department, is greatly appreciated.

I would like to show my gratitude towards the members of the examination board who where able to make time to review the results of this research and where willing to participate in this process.

Special thanks go to Samuel Torreborre, the patients who volunteered during the measurements at the LUMC and the LUMC personnel who where friendly enough to support and tolerate my presence when working in their midits. I would also like to use this opportunity to thank Harry Jansen and Patrick van Holst for being able to squeeze the construction of a very nifty measurement frame in their busy schedule. Jos van Driel was kind enough to help me out with several details involving my measurement setup, and was always able to make time to help me debug or solve my software problems. Also Rob Luttjeboer’s help by arranging a suitable measurement location is greatly appreciated.
Finally I would like to thank my family and housemates for their support and understanding along the road. I especially would like to thank my fellow students with who I shared the same master office, for their inspiring companionship, sincere interest and sufficient cosiness to create a very pleasant working atmosphere.

Teunis Schuurman,
Delft,
May 16, 2011
Summary

The human abdominal cavity is an interesting part of the human body, containing a lot of important organs which, due to relative soft boundaries, are quite accessible for examination. Conditions like mechanical bowel obstruction, obstruction of the inferior vena cava, abnormalities of the viscera and existence of abdominal aortic aneurysms can be detected using physical examination. This technique involves ‘feeling’ for abnormal mechanical behavior, unusual masses or pulsation sources, and is basically a manual assessment of mechanical properties and excitation sources. Changes in mechanical properties change the feel, but also changes the dynamical properties of the abdominal system. As such, it can be expected that changes in mechanical properties or changes in excitation sources should also cause an alteration in system dynamics, and should be visible as a change in vibration pattern of the abdomen. This provides an quantifiable parameter which can be measured and monitored. Previous research showed that it was possible to use a Laser Doppler Vibrometer to detect vibration patterns on the skin surface above the carotid artery and the thoracic cage, and it is assumed that this approach can also be applied in a scan of the abdominal surface.

A measurement setup is designed to measure the cardiac induced vibration of the abdomen. An ECG monitor registers heart activity and is used to synchronize the measurement with the QRS complex of the heart. An commercial Laser Doppler Vibrometer is used to measure the vibration on a grid consisting out of several dozen of reflective stickers placed on the abdomen surface. By using the QRS complex as a trigger, a sequential scan passes along the defined gridpoints and measures their velocity over a sample period of a few seconds. Afterwards the sequential data is used to generate an approximate parallel scan of the surface. By including a Finometer into the setup it is possible to monitor the brachial blood pressure waveform and to define frequency response functions between the input blood pressure and output skin vibrations.

The measurements reveal that there is indeed a repeatable vibration pattern present at the abdomen, which synchronizes with heart rate. Although the velocity of this vibration pattern is affected by respiration, its effect can be reduced by applying averaging over multiple scans, making it is possible to obtain a clear pattern which synchronizes with the cardiac cycle and is most likely generated by pulsation in the abdominal aorta and possibly by other arteries. It is also observed that the vibration pattern differs in shape depending on location on the abdominal surface. Gridpoints located above the abdominal aorta show a clear cardiac pulse, even appearing to show the wave velocity of the pressure pulse, while gridpoints positioned further away show an increased complex signal and a significant reduction of vibration amplitude.
Several different data processing techniques are tested. It is found that the vibration patterns show quite some detail, and it is necessary to discriminate between measurement location. Frequency response functions between bloodpressure waveform and skin vibrations are defined and appear to show the presence of dynamical behavior. Accompanying coherence functions show quite high values of coherence up to 10 Hz, but this tends to vary between measurements. This suggests that, although maybe not on all occasions, abdomen vibration is primarily caused by blood pressure waveform.
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Chapter 1

Introduction

1.1 Background

The human abdomen is the largest body cavity in humans and contains a large number of important organs. It contains the majority of the digestive track, the liver, kidneys, the spleen, pancreas and several other important organs. In contrast to other cavities, the abdominal cavity is only partially covered by bones and bone-like structures, making the organs it contains quite accessible. Most of its internal organs, the so-called viscera, are only supported by membranes, allowing the viscera to move within the abdominal cavity. Due to this ‘free’ structure, it is possible to evaluate its condition by means of physical examination, which partially focuses on the detection of unusual masses, rigidities, alterations in sliding friction between the membranes (rubbing) and the existence of pulsatile masses. This method can be used to reveal mechanical bowel obstruction, obstructions in the inferior vena cava, abnormalities in most abdominal organs and the existence of abdominal aortic aneurysms (AAA’s) [Ferguson 1990]. This examination method requires a certain degree of training, especially when searching for specific abnormalities, as Fink et al. [2000] demonstrated for the detection of AAA’s. Another disadvantage is the need of physical contact which, especially during deep palpation, can be too painful for a patient to bare, or which, due to the condition of the patient (for instance burn victims or patients which where exposed to hazardous toxins), should be avoided.

Physical examination is based on the effect of changes in mechanical properties (rigidities, sliding friction) and/or dynamic properties (unusual masses, pulsation sources, etc) on the ‘feel’ experienced by the physician. In a similar way as these properties effect the perceived feel, the same processes should also effect the motion within the abdomen, which can be quantified in measurable properties. Assuming that the motion occurring within the abdomen is transferred to the overlying abdominal wall and skin, measuring the motion of the abdominal wall should prove an indication of the internal abdomen behavior. This approach could be used to identify changes of internal excitation sources or could be used to indicate changes of the mechanical path between excitation source and measurement location.

There are several methods available to detect the motion patterns of the abdomen surface, like Laser Doppler Vibrometry (LDV), photogrammetry [Flora et al. 2001], holographic interferometry [Hok et al. 1978], accelerometers and several other exotic techniques. Due to the fast setup, large range of operation, and relative ease of operation, LDV is considered to be the most practical within this application, focusing this research on the techniques suitability to detect and monitor condition-specific vibrations.

LDV is a non-contact measurement tool based on the principle that laser light experiences a
Doppler shift in its wave frequency when reflected by a moving object, similar to what is observed for sound, but due to the high frequencies involved, not visually observable. By demodulation of the signal it is possible to detect small displacements (nm) and velocities (µm/s) at a wide range of frequencies (from 0 Hz up to 1.5 MHz), making it a perfect tool to detect vibration and to scan for vibration patterns.

Although mostly used in mechanical applications, there is also interest from the medical sector to apply this approach as a non-contact examination method. Possibly inspired by Dr. McCoy’s biobed, as envisioned in the SciFi serie Star Trek, effort is made to develop non-contact diagnostic tools. By detection the vibration of the skin, it is possible to extract the blood pressure waveform of underlying arteries. Possible applications range from heart beat monitoring of infants or burn victims [Scalise et al. [2004]] to long-range triage of wounded soldiers [Desjardins et al. [2007]].

It is also possible to apply LDV as a scan, (as suggested in Hong and Fox [1994]) measuring the vibration of a large skin surface, for instance the abdominal wall. When performed with sufficient accuracy, the skin vibration patterns could reveal information about the condition of underlying abdominal organs. Such a scan would only require a small amount of preparation, will be relatively fast and cheap, and could be performed without the need of specialist training. Automated data processing could alarm the physician when abnormal motion patterns are observed, triggered by abnormalities within the abdomen internals, making it usable as a pre-scan tool.

1.2 Problem description

As mentioned in the previous section, the measurement of skin vibration and detection of ‘abnormal’ vibration patterns could reveal the state of the abdomen internals. Several techniques are available to detect skin surface motion, the most suitable techniques to detect surface vibrations are treated below. Laser Doppler Interferometry is based on the property that a coherent laser light beam, intercepted and reflected by a moving object, will experience a Doppler shift in its frequency, similar to the Doppler shift experienced in acoustics. This frequency shift can be used to accurately measure the velocity or displacement component of the object in the direction of the laser beam. Accuracies in the order of nanometers or micrometers per second are achievable by this fast, non-contact acquisition method. By introducing multiple scanners in the setup, it is possible to measure 3D surface. An important disadvantage of this technique is its inability to
measure multiple points in parallel, however this can be resolved by using a multibeam scanner [MetroLaser, Inc. Polytec CD-ROM].

Electronic Speckle Pattern Interferometry (ESPI) measures the displacement between two laser pulses. An image of the current speckle pattern is subtracted from a reference image and the resulting correlation fringe patterns reveal the displacement between the two images. Images are obtained by means of a CCD (digital image sensor), which limits the lateral resolution obtainable. In order to obtain speckles large enough to be detected by the CCD, an optical rough surface is necessary. ESPI system typically require a Class 4 laser, yielding the need of special safety precautions. All surface points are measured simultaneously, but each frequency requires at least 3 shots in order to be detected [Spagnolo and Ambrosini 2010, Polytec CD-ROM].

Accelerometers measure the dynamic displacements of a mass in a simple mass spring system and calculate the acceleration by determining the displacement of the mass. Displacement can be measured by means of capacitive, inductive or piezoelectric principles. The measurement principle is based on acceleration (inertia) effects, and becomes more sensitive at increasing values (and frequency). As such, resolution at lower frequencies is somewhat limited (figure 1.2). The use of accelerometers brings advantages like its ability to perform a simultaneous measurement of a multitude of points, lower investment (for low number of points) and makes it possible to measure points which are normally hidden from sight. Important disadvantages are the time consuming setup, high costs when large numbers of points are measured and the existence of several error sources (glue interface, alignment, calibration, mass loading effects on surface) which require expertise in order to be kept to a minimum. [Polytec CD-ROM].

Photogrammetry compares video images from two (or more) cameras to measure the position of the markers. Specific resolutions are quite low, resolutions ranging from 3 micrometer to 50 micrometer are reported, but highly dependent on vibration frequency (figure 1.2). (A typical commercial system can achieve a resolution up to 1 micrometer). An important advantage of this technique is its ability to measure a large amount of markers, positioned on 3D surfaces, stretching up to several square meters within seconds. Dependent on the system used, the frame rate can be in the order of several kHz. A disadvantage is the necessity to place markers on the surface, which can be a time consuming task. [GOM GmbH].

Comparing the properties of before mentioned measurement techniques, it can be concluded that LDV combines easy setup with the ability to cover a wide range of velocity amplitudes and vibration frequencies. These favorable properties make it an interesting tool to be used in the detection of skin vibration. Currently only limited experience has been reported considering the usage of this scanning technique to detect vibration patterns occurring on large biological surfaces. The goal of this research is to evaluate the usability of Laser Doppler Interferometry to track abdominal skin vibration and the ability of this approach to evaluate vibration excitation sources and the mechanical properties of the transmission path between source and measurement position.

Usability will be quantified by several requirements considering the ability to relate the measured signal to distinct sources, the repeatability of the measurements results and the ease of implementation. This will be achieved by studying literature and testing and refinement of the method during experiments.

1.3 Related work

Laser Doppler Interferometry is already a common tool in the clinical medical field, but current commercially available applications are mostly limited to blood perfusion measurements or to detect micro circulation within biological tissue [Oxford Optronix Ltd., Perimed AB]. However, the usage of Laser Doppler Vibrometers to accurately measure vibrations in mechanical systems has triggered several research groups to incorporate similar systems as an tool in their
research. Over the years a wide variety of biomechanical applications have been published, ranging from detection of the mechanical response of the eardrum to sound [Huber et al. [2001]], detecting changes in the mechanical properties of skin tissue [Takei et al. [2004]] or to determine the amount of arteriovenous fistular blood flow after surgical construction [Aoki et al. [1996]].

Several interesting studies, related to the context of this study, have been focused on using Laser Doppler Interferometry as a non-contact measurement of heartbeat and blood pressure waveform. [Hong and Fox [1993a]] appear to be the first to suggest the use of Laser Doppler Vibrometry as a non-contact technique to use skin vibrations to detect the blood pressure waveform of underlying arteries. Their article describes an ‘optical stethoscope’, based on laser Doppler vibrometry, and confirmed the ability of the instrument to measure heartbeat frequency and its ability to detect changes in the pressure waveform inside an artery embedded in tissue. They also suggest the possibility to perform such measurements ranging over large body surfaces by means of scanning [Hong and Fox [1993b], [Hong and Fox [1994a], [Hong and Fox [1994b], [Hong and Fox [1997]].

In Europe, Scalise et al. [2004] used a commercial (Polytec) laser vibrometer for non-contact heartbeat monitoring. In contrast to the work of Hong et al., their goal is to obtain a ‘remote’ heartbeat monitor, for instance to be used within a MRI scanner (where the existence of powerful magnetic fields can cause dangerous interaction with the wiring of an ECG recorder) or in case direct contact is dangerous (infants, burn victims, secondary exposure of medical personal to toxic material, etc). Their research focused on measuring the thorax movement, although some experi-
ments have been performed on skin surface above the carotid artery, and was primary focused on validating the equivalency between ECG (Electrocardiograph) and VCG (Vibrocardiograph). This was done by comparison of the time signals, their frequency content and by comparison of Heart Rate Variability (HRV) indicators of both signals [Scalise et al. 2006, Scalise et al. 2008, De Melis et al. 2007, Scalise et al. 2008, Scalise et al. 2010]. Desjardins et al. [2007] reports the use of LDV as a method to remotely measure the blood pressure waveform within the human carotid artery over large ranges. This research is performed in cooperation with the US Naval Undersea Warfare Department and is accompanied by the assignment of a patent for the method of using a laser-based interferometer as a non-contact blood pressure waveform detector [Antonelli et al. 2006]. Other possible detection methods for skin vibration include the use of laser triangulation [Wu et al. 2007] in which skin motion is detected by the displacement of the reflected laser beam over a CMOS image sensor, Hast et al. 2002a used a self-mixing laser Doppler interferometer, only consisting out of a modified laser diode, to record the blood pressure pulsation profiles in a non-invasive manner [Hast et al. 2001, Hast et al. 2002b] and [Hast et al. 2006]. Hök et al. 1978 report the application of laser holographic interferometry to detect chest-surface movements induced by heart action and its possibilities to detect abnormal heart functioning.

1.4 Project organization

Supervisors:
Prof. dr. ir. D.J. Rixen, Delft University of Technology;
Dr. J.W. Hinnen, LUMC Surgery
Dr. ir. C.A. Swenne, LUMC Associate Professor Cardiovascular Physiology

1.5 Disposition

Chapter 2 will outline the problem statement and define the research questions on which the research will be based. It also contains a brief insight into the proposed measurement technique and some of the problems which were expected to be encountered before the start of the research.

Chapter 3 contains the results of a feasibility study, which consists out of a literature study of similar setups, the techniques previously applied and experienced obtained. It also contains a description and results of a several experiments which have been performed to assess the feasibility of LDV measurement in detecting skin vibrations.

Chapter 4 presents the results of an intense literature study of expected motion sources contained in the abdominal cavity and relevant analysis methods used to evaluate them. Section 4.3 consists out of a thorough description of the measurement setup and the protocol used during the research. Finally section 4.4 treats the results obtained during the experiments which where performed to test LDV settings and optimize the setup.

Chapter 5 summarizes different analysis techniques that have been used along the research project and evaluates their advantages and disadvantages when applied on this subject.

Chapter 6 presents the results of some of the different physiological conditions on skin vibration, like underlying tissue, respiration, stomach content, tensioning of abdominal muscles and presence of an abdominal aortic aneurysm.
Finally chapter 7 presents the conclusions of the research, chapter 8 discusses future work and recommendations for further research.

1.6 Abbreviations

A short list of regularly used abbreviations:

**LDI** Laser Doppler Interferometry.

**LDV** Laser Doppler Vibrometry or Laser Doppler Vibrometer.

**ECG** Electrocardiograph.

**VCG** Vibrocardiograph.

**LUMC** Leiden University Medical Center.

**AAA** Abdominal Aortic Aneurysm.

**EVAR** Endovascular Aneurysm Repair.

**(F)FT** (Fast) Fourier Transform.

**iFFT** inverse Fast Fourier Transform
Research objective and investigation method

2.1 Problem statement

The goal of this research is to evaluate the possibility of employing Laser Doppler Vibrometry (LDV) as a diagnostic tool to detect physiological changes within the abdomen. It is assumed that some physiological changes within the abdomen result in change of mechanical properties, which would affect internal dynamic behavior. Such a change in dynamic behavior should be detectable to the outside due to a presumed influence on abdominal skin motion, which can be registered by means of a laser Doppler vibrometer.

![Figure 2.1: Schematic of detection system signal transfer. It identifies the three major information transfer mechanisms: 1. body dynamics describing the relation between changes in abdomen mechanics and skin motion; 2. signal transfer transferring skin motion to the laser vibrometer due to light transmission; 3. Signal processing of the signal recorded by the detector.](image)

Suitability should be quantified according to the methods ability to register change in skin motion as a response to a physiological change. Referring to figure 2.1 which shows a schematic of the system, it is possible to identify an information transfer route:

(a) Excitation sources (heart beat, pulsation of arteries, blood flow patterns, breathing, motion...
of viscera) excite the internal mechanics of the abdomen, causing vibration of the surrounding abdominal tissue.

(b) Vibrations are transferred from the excitation source through the abdomen to the abdominal wall and overlying skin. Mechanical properties (like stiffness, damping, mass distribution) will influence this transfer. As such, the vibrations of the abdomen surface contain both information of the excitation source as well information regarding the mechanical properties of the path between source and measurement location.

(c) The vibration signal of the skin is measured by means of a LDV system, and subject to interference occurring during reflection and along the optical path.

(d) Signal conditioning. In order to obtain reliable, meaningful results, it is necessary to condition the signal into a representation suitable for use in clinical settings. This condition influences the final results.

This research will primarily focus on the origin of the motion (part (a)) and investigate the technical particulars of this scan (routes (c) & (d)). Efforts spent on modeling the internal mechanics describing the transmission from 'source' to skin motion are extremely limited. Although this internal transfer is considered vital to the overall suitability of the method, a correct model would require detailed information of in vivo tissue mechanics, the validation of the skin response as a function of internal changes in dynamics, etc. To current knowledge, such a rigorous model has not been documented in literature yet, and will most certainly fall outside the current knowledge of the author. Due to this lack of expertise, and even more importantly, an unavailability of the resources required to record internal mechanics and material properties, it was chosen to shift the focus towards the detection side of the method, quantifying possible interferences and developing proper scanning protocols and postprocessing.

### 2.2 Research questions

After carefully consultation of literature and evaluation of the available options, it was decided to concentrate the research on the application of Laser Doppler Vibrometry to detect abdomen surface vibration. The research question has been defined as followed:

*Is Laser Doppler Vibrometry able to detect mechanical responses caused by physiological events occurring within the abdomen?*

In order to answer above problem statement, it will be divided into several research questions, which will be separately answered over the course of the research.

**Skin reflection** Does the human skin reflect sufficient optical signal back to the LDV detector to register skin motion? How do these so called skin optics work and how can they be improved?

**Signal noise** Which processes and mechanisms interfere with the signal? What is their influence on the measurement signal and can they be isolated and/or compensated?

**Signal content** What is the content of the measured signal. Is it possible to identify periodic signals of skin motion?

**Repeatability** How repeatable are the measurements? What are the major (biological) causes of spread in the data and how should the protocol be adapted in order to minimize this spread?
Data processing  How should the measured data be processed in order to maximize usability of this method?

Data interpretation  How does, at a basic level, a physiological change in abdominal internals influence the skin motion?

2.3 Measurement setup

Most of the measurements within this research are performed by using a commercial Polytec system, composed out of a PSV-I-400 scanning head, a PS-E-401-FG1 junction box and an OFV-5000 controller. This setup makes it possible to scan large areas and record displacements and velocities in the direction of the laser beam with nanometer resolutions and frequencies up to several hundreds of kHz.

It is assumed that the cardiac cycle (about 1 Hz) is a dominant periodic excitation signal and the heartbeat will be used as a trigger to switch between scan points assigned on the surface of interest. This approach results in scans of multiple points, all experiencing similar excitation. As a result, an approximation of a parallel scan is obtained, although the setup is only fast enough to perform sequential scanning (figure 2.2). Heartbeat will be monitored by a commercial electrocardiograph (ECG) recorder, which will trigger the scan at the QRS complex of the heartbeat.

![Figure 2.2: Parallel approximation obtained by sequential scanning. Point 1 is scanned on time t = 1, point 2 is scanned at time t = 2, etc. Assuming that the motion is primary the result of an input signal which repeats itself at point 1, 2, 3, 4, the surface should be similar at all four instants. This means that the sequential measurements can approximate a parallel scan over the grid.](image)

Another possible option is to analyze the system response between the input source and output. By measuring the blood pressure waveform signal (input excitation) and simultaneously recording the surface vibration (output response), it could be possible to define a transfer function between both.

An important part of this research would consist out of evaluating the influence of ‘biological’ disturbances. It is assumed that the cardiac pulse would be the main internal ‘resonator’, but it does not vibrate on one single position. The heart itself will act as a pulse location, but it can also be expected that other arteries, like the abdominal aorta, will also act as a pulse initiation site. Similarly, although at lower frequency, respiratory motion and motions of the digestive track will also act as pulse initiation sites and will influence skin motion (figure 2.3).
Figure 2.3: Internal motion initiators: The heart and the arteries will pulsate during the cardiac cycle [(a)], but also the respiratory motion [(b)] and the digestive track [(c)] will act as motion initiation sites.
Chapter 3

Feasibility experiments of concept

3.1 Introduction

In the early stages of this project it was still unknown if the available Laser Doppler Vibrometer would be able to detect the motion of human tissue and if the measurements would be repeatable. As such, it was decided to perform some initial experiments in order to quickly assess the feasibility of this novel approach. Encouraged by the results of [Desjardins et al. 2007] and [Hong and Fox 1993b], several tests were performed trying to obtain similar results as stated in literature. All of these experiments were performed on the wrist of the hand, if not stated otherwise.

3.2 Results obtained from literature

3.2.1 Related research

An review of literature revealed that the use of Laser Doppler Vibrometry to detect skin vibration has successfully been implemented in the past.

Hong and Fox [Hong and Fox 1993a] & [Hong and Fox 1997] developed a non-invasive optical method to detect cardiovascular pulsation, using a custom designed ‘optical stethoscope’ based on Doppler Interferometry (figure 3.1(a)). Using this system they, among other things, were able to measure the vibration of the thorax near the cardiac apex (bottom of the heart which tends to ‘tap’ the chest wall, usually palpable [Mediscuss - Exam. of the Card. Apex Beat]). Their time domain measurements showed the presence of the cardiac pulse within the signal [Hong and Fox 1993b]. Other experiments [Hong and Fox 1994a] showed that the complexity of the signal is dependent on measurement location: measurements at the radial artery revealed a relative simple, velocity profile, resembling an arterial pressure waveform while the profile obtained on the chest wall showed a more complex shape. In [Hong and Fox 1994b] they have build a physical model of a artificial flexible artery embedded in artificial tissue. Their optical stethoscope was used to record tissue surface vibration and it was possible to theoretically predict skin vibration caused by the blood pressure waveform.

[Desjardins et al. 2007] used a commercial Polytec system (very similar to the one used during this research) and tried to develop a remote and non-contact method for obtaining the blood-pulse waveform of the carotid artery (figure 3.1(b)). Although not yet able to correlate the measured velocity profiles with the pressure wave form, they did correlate the results with ECG recordings and were able to identify the systole and diastole patterns within the velocity profile.
The measurements are usually applied on bare skin, although they sometimes applied a reflective sticker to improve signal quality.

![Helium Neon Laser: 633 nm, 5 mW](image)

Detector Head
Transimpedance Amplifier
Fiber bundle
Stethoscope Bell
Skin Vibration
Skin Layers
Artery under skin surface

Figure 3.1: An examples of setups used in literature: Figure 3.1(a) shows the optical interferometry setup used by Hong & Fox [from Hong and Fox [1997]]. A stethoscope bell is used to ensure a fixed relative position between detector and skin surface. Figure 3.1(b) shows the setup described by Desjardins and Antonelli, which measures the vibration above the carotid artery using a commercial system from Polytec [adapted from NUWC usnavyresearch].

Both Hong & Fox and Desjardins addressed the effect of skin optics on the signal quality and mention their efforts taken to improve the quality of the reflected laser beam. Visible light is not reflected at the air - skin interface, but mostly penetrates a certain depth within the skin layers before being backscattered out of the skin. The precise mechanism is quite complex, dependent on the wavelength of light and the composition of the biological tissue. Models that predict the reflection and transmission do exist, but their accuracy is highly dependent on the ability to characterize the optical properties of the tissue.

Hong and Fox [1997] claim that that blocking the light penetration (with an optically opaque film, blocking by painted film at the point of laser focus on the film or by direct painting of the skin surface) improves signal. Desjardins used a retro-reflective sticker at some instances to improve signal intensity of the reflected signal, while preventing optical transmission into the skin. However, both agree that, especially at areas with little hair growth (hair follicles tent to absorb and scatter light), little skin preparation is needed to obtain sufficient signal.

3.2.2 Skin Optics

Light hits the skin surface and is partially absorbed and partially redirected towards the detector. However, it should be realized that skin is a biological tissue, and as a result, shows quite complex optical properties. Skin is composed out of several distinct layers, all having different optical properties. Figure 3.2 shows a cross-section of the skin, identifying the following layers: Stratum Corneum, The outermost layer of the skin, composed out of flat, keratin filled plate-like envelopes, consisting out of dead cell originating from the epidermis. This tissue shows no light reflection, transmission through the layer is approximately uniform for visible light. Only a small fraction (4 % to 7 %) of the incident light is reflected due to the change in refrac-
Epidermis The epidermis, combined with the stratum corneum, forms the most outer skin layer. The thickness of the stratum corneum and epidermis is still small enough such that its contribution to remittance is minimal over the entire visible and near infra-red spectral regions. Epidermis is thin enough such that its scattering properties can be neglected compared with the dermis. It is mostly transmitting or absorbing, most dominantly absorbing for small wavelengths (UV light).

Dermis Scattering is important in the dermis. Longer wavelengths penetrate deeper into the skin than shorter wavelengths (on UV-visible-near infrared radiation). Light of 600 - 1300 nm penetrates quite deep into skin and human tissue in general (the so called 'optical window'). Total absorption coefficient within the dermis is quite dependent of hemoglobin absorption [Jacques and Salerud 1998].

As mentioned above, skin remits visible light mainly due to scattering within the skin layers, causing a diffuse, but quite light appearance. Light above 600 nm, like the 630 nm used in the commercial Polytec laser [Polytec Manual], is mostly scattered instead of absorbed. It also penetrates deeper into the skin (it can be stated that longer wavelengths result in deeper penetration), up into the dermis. Additional details regarding the optical properties can be found in appendix B.

3.3 Experimental Studies

Several tests where performed to evaluate the impact of skin reflectance on signal quality. In order to test this, three separate reflective conditions where tried on the skin of the inside of the wrist.

- Bare skin: unconditioned skin on the inside of the wrist (no skin hair present);
- Reflective sticker: a reflective sticker applied on the skin;
- Oily skin: skin oiled with baby oil in order to increase reflectivity.

Measurement are performed with the subject holding his breath, the LDV settings are summarized in the table below.
### Table 3.1: Settings used during the FFT measurements. LP = Low Pass filter, VD = Velocity Decoder.

3.4 Results of Experiments

No significant difference could be observed between the investigated skin treatments. Comparison of the measured frequency spectra for the separate treatment did not show very different results. Furthermore the variation appears quite similar!

Comparing the frequency spectra obtained by the measurements revealed the presence of a peak around 1 Hz, which coincides with the to be expected heartbeat frequency. A later experiment incorporated the use of an ECG recorder (appendix C.1) proving that this peak indeed coincides with the heartbeat frequency. Figure 3.3 shows that both the LDV and the ECG signal show a similar frequency peak, even if the heartbeat frequency has been raised due to physical exercise. Physical exercise seems to increase the amplitude of the heartbeat peak, possibly the result of the baroflex reflex.

In order to simulate the results obtained by Desjardins et al. [2007], the LDV has been targeted to the carotid artery of a volunteer. The same LDV settings as defined in table 3.1 were used, although this time no ECG signal was logged and the acquisition setting where altered in order to apply a Hann window on the sample window. Figure 3.4 shows that the LDV signal gives a very clear peak at the to be expected heartbeat frequencies. Physical exercise results in a larger frequency, coinciding with an increased heartbeat frequency.

3.5 Conclusions

Both literature and the preliminary test results seem to suggest that sufficient signal quality can be obtained without the use of reflective stickers. Light, especially the 630 nm wavelength laser light of the LDV penetrates into the skin, until backscattered in the dermis. However, as Desjardins et al. [2007] mentions, the presence of hair follicles and other micro structures within the skin could cause irregular scattering, which will be visible as noise within the signal.

The preliminary experiments did not show any significant difference in the resulting magnitude spectra when comparing measurements performed on bare skin, skin treated by oil and when a retro-reflective sticker is applied. The spectra show similar frequency peaks and the variation between the separate skin conditions is roughly similar. This appears to indicate that bare skin gives enough backscatter to provide sufficient signal, but it is also possible that the motion of this skin sample shows such a large variation that any variation associated with skin preparation is not noticed. Although it is not know if the presence of the to be expected hair and hair follicles on the abdomen surface will significantly affect signal quality, it is at least shown that the application of retro-reflective stickers yield similar results as on bare, bald skin.

The measurements suggest a direct relation between skin motion at the wrist and the heart beat frequency. Increasing the heartbeat frequency results in an increase in frequency of the peaks observed in the frequency spectra. This observation is in agreement with the results reported by
**Figure 3.3:** Averaged frequency spectra of the bare skin and the ECG recording after performing physical exercise and after some rest. Intervals (mean $\pm 2 \times \sigma$) have been included to indicate the standard deviation present within 5 acquisitions used in each dataset. Both during raised HB frequency (figure 3.3(a)) and after rest (figure 3.3(b)) the first peak in magnitude of the skin velocity signal coincides with the Heartbeat frequency registered by the ECG recording, revealing the relation of the cardiac cycle on skin motion.

**Figure 3.4:** Frequency spectrum obtained by pointing the vibrometer on the carotid artery. Measurements where performed on bare skin, without any special preconditioning. In contrast to the averaged spectra previously shown, these spectra are obtained from single acquisition. Subfigure 3.4(a) shows the spectrum with the subject in rest, subfigure 3.4(b) shows the spectrum after the subject performed some physical exercise. A (medium-quality) low pass filter was incorporated with a cutoff frequency of 50 Hz. The first peak appears to coincide with the heartbeat frequency and increases in frequency when the heartbeat frequency is raised.

Hong & Hox and Desjardins et al. 

Very similar to the results reported in Desjardins et al. [2007], it was indeed possible to extract heart beat frequency by measuring the motion of the carotid artery. The resulting amplitude spectra and time signal (figures 3.4 and 3.5) show a periodic contribution which is believed to be
coincide with the heartbeat frequency.

Although no effort has yet been taken to determine the motion at the abdomen surface, it is believed the abdominal surface should show similar behavior as observed on the wrist and carotid artery: e.g. the abdominal wall should vibrate in response to cardiac activity. The results of both literature study and the performed measurements showed that the LDV should be capable to detect such motion.

Figure 3.5: Figure 3.5(a) shows the time signal of the skin velocity obtained by pointing the vibrometer on the carotid artery. The skin vibration shows a clear periodic signal. A (medium-quality) low pass filter was incorporated with a cutoff frequency of 50 Hz. Figure 3.5(b) shows the velocity profile in more detail.
Chapter 4

Performance assessment of proposed measurement setup

4.1 Introduction

After the first initial experiments it was concluded that the available Laser Doppler Vibrometer was indeed able to detect relevant motion patterns of the skin, was able to obtain similar results as specified in the articles of Desjardins et al. [2007] and the research of Hong and Fox [Hong and Fox 1993a till Hong and Fox 1997] and should be able to obtain sufficient performance without the explicit need of retro-reflectors. Furthermore the experience revealed that implementation of a LDV within the measurement protocol should not prove to be to difficult.

The experiments confirmed the expectation that the cardiac cycle would be a fairly dominant motion source and after obtaining these encouraging results, effort was taken to design the measurement setup and literature was studied in order to obtain more knowledge regarding the to be expected motion sources of the abdomen, to determine the magnitude of disturbance that could be expected due to the optical properties of the skin and the practical usage of the LDV in general.

Using the experience obtained by the preliminary test and the knowledge obtained from literature, a measurement setup was designed, which would be able to detect abdomen surface motion. The experiments performed within this research phase where meant as preliminary tests, implemented to assess the influence of different acquisition setting and measurement protocols, instead of attempts to gather data. MATLAB algorithms where written and tested on the obtained results and there was some experimentation regarding post-processing of the results.

4.2 Results obtained from literature

4.2.1 Introduction

This section summarizes the results of a literature review, which was performed in order to determine the to be expected influence of organ motion on skin motion. It appears to be necessary to make a subdivision between organ motion and organ position. Organ motion will be defined as relative fast change in organ position (timescale of within minutes), organ position will be defined as slow or ‘static’ changes in position, for example due to change in posture or quasi-static processes like bladder and rectum filling [Langen and Jones 2001].
Figure 4.1 shows a schematic overview of the abdominal surface and the (approximate) position of the major abdominal organs (viscera). All of the viscera are contained in the peritoneum, a membrane that covers the inside wall of the cavity (parietal peritoneum) but also every organ or structure contained in it (visceral peritoneum). The space between the visceral and parietal peritoneum, the peritoneal cavity, normally contains a small amount of serous fluid that permits free movement of the viscera, particularly of the gastrointestinal tract, contained inside the peritoneal cavity. [Encyclopedia Britannica].

**Figure 4.1:** Schematic overview of the abdominal surface and the relative position of the major abdominal organs. Figure 4.1(a) shows the outer landmarks and the viscera below them. Figure 4.1(a) shows the major viscera within the abdominal cavity. Figure 4.1(c) focuses on the digestive track in the LDV scan region of interest. Figure 4.1(d) details on the abdominal aorta and other parts connected with the cardiac system.

**Organ position**

Organs can shift in position due to change in posture. This organ mobility is highly dependent on the physical structure containing the organs. For instance, organs within the skull hardly change in location (less than 2 mm), while organs contained in the abdominal cavity are highly sensitive to changes in posture. Furthermore organs change in thickness and shape depending on posture.
Furthermore, organ position can change on a day-to-day level, due to weight changes, digestive system content or bladder content. This type of position change, commonly referred to as inter-fraction motion in the field of radiation treatment, is mainly associated with parts of the digestive system. It can be highly dependent on rectum and bladder content. This kind of position change is a dominant source of change in location for the organs contained within the pelvic area, like the gynecological organs, prostate, seminal vesicles, bladder and rectum, which appear to show only limited motion as a response to respiration [Langen and Jones 2001]. Although also influenced by the same sources which cause interfraction motion, the abdominal organs are much more influenced by the faster-scale organ motion. Organ position will most likely not be a matter of concern when comparing the result contained within a single surface scan or even measurement session. However, it is expected that these phenomena will have to be included when comparing results between separate measurement sessions (inter-patient measurements).

Organ motion

Organ motion, commonly referred as intrafraction motion within the field of radiation treatment, is associated with faster changes in organ position. Organ motion is usually caused by respiration or the result of cardiac motion, and is a dominant type of motion for most of the abdominal organs like the liver, stomach, diaphragm, pancreas, kidneys and small and large intestines. The digestive track is also capable of creating motion by means of peristaltic action. The time scale of this type of organ motion small enough to be visible within the measurement period of single surface scans.

Additional effort will be made to determine the influence of the following sources of organ motion occurring within the abdomen:

- Pulsation of the cardiac cycle
- Motion due to respiratory cycle
- Vibrations caused by gastrointestinal motion

Figure 4.2: Definition of abdominal directions during the measurements and commonly used anatomical definitions. The red coordinate system defines the coordinate system used in the measurement results.
Pulsation of the cardiac cycle

The pumping action of the heart will result in a periodic change of pressure through the arteries. Arteries close to the heart, like the abdominal aorta, are quite elastic and show a significant amount of dilation due to the pressure variations. Dependent on age and sex, the systolic diameter increases 2% to 20% (5% to 25% in a female population) larger compared with the diastolic diameter [Son [1993]]. However, it should be noted that increased age results in larger and stiffer arteries, resulting smaller diameter changes between systole and diastole [Lan [1992]].

A more direct path is that the dilation and compression of the heart transfers mechanical and acoustic energy to the chest surface. However, the vibration from acoustic waves appears to be small compared with the bulk mechanical displacements [Hong and Fox [1994a]].

Arteries expand and retract under influence of the arterial pressure. The shape and magnitude of the blood pressure waveform is a (complex) function of arterial wall stiffness, location within the arterial tree, respiration rate, local blood flow and other factors. Speed and amplitude of the pressure wave increases when traveling from the large, elastic arteries into the smaller, more muscular vessels due do the decrease in vessel compliance. This change results in a significant distortion of the pressure contour: the systole part of the waveform becomes narrower and increases in magnitude, the small pressure drop of the dicrotic notch is damped, eventually disappears and replaced a hump in pressure. This tendency to damp high frequency components of the waveform is attributed to the viscoelastic properties of the arterial wall.

Besides this dampening effect, arterial stiffness also influences the reflection pattern. During each heart cycle, a forward-traveling pressure wave travels from the heart through the arterial tree. Bifurcations and the periphery cause a reflection of the pressure wave, the backward-traveling wave. When the forward and backward-traveling waves meet, this causes an locally varying increases the peaking of the pressure wave. In a young (elastic) arterial tree there exists a pronounced difference in central and peripheral pressures: systolic pressure increases further away in the tree, while diastolic pressure remains essentially unchanged, basically causing an amplification of blood pressure waveform.

Aging of the arterial tree is accompanied with increased wall stiffness, and reduces the difference between central and peripheral systolic pressure, while increasing pulse pressure (as in the aorta). This is caused by stiffer arteries, which transmit the pressure pulse wave with a higher wave velocity. This results in a larger than normal back-traveling pressure wave, returning faster and increasing pressure of the late systolic peak pressure. Peripheral systolic pressure also reduces, reducing the difference between central and peripheral pressure (figure [4.3] [Wilson et al. 2001]).

The blood pressure waveform is also influenced by the respiration cycle (the motion of the diaphragm influences heath filling which induces variations in blood pressure), rate of pumping (increased output results in increased blood pressure), total system resistance and blood viscosity. The influence of respiration on heart filling and blood pressure also causes variation in heart beat frequency, known as Respiratory sinus arrhythmia (RSA), and caused by feedback of the baroreflex. RSA coincides with respiration rate and is very pronounced in children or adults with excellent cardiovascular health like endurance athletes. RSA becomes less prominent with age, diabetes and cardiovascular disease [Wikipedia Baroreflex [Wikipedia Respiratory sinusarrhythmia]].

This phenomenon will influence cause a significant of variation in QRS - QRS wave times, and as such, effect triggering.

Figure [4.5] shows an example of an ECG track showing this occurring in a healthy young subject.

Sounds generated by the cardiac cycle are usually related to the 0.5 - 50 Hz bandwidth.

Motion due to the respiratory cycle

Motion caused by the respiration can be split in motion caused by the respiration cycle itself and vibrations caused by respiration (e.g. respiratory sounds).
Figure 4.3: Central pulse pressure waveform. Systolic and diastolic pressures are the peak and trough of the waveform. The augmentation pressure is the additional pressure added to the forward wave by the reflected wave. The dicrotic notch represents closure of the aortic valve. Time to wave reflection is defined from the point of rise in the initial ejection wave to the onset of the reflected wave. The reflected wave in this central pressure waveform results in increase in pressure during systolic flow. Adapted from Nelson et al. [2010].

Organ motion caused by the respiration cycle is of particular interest in the field of radiation treatment, because it can cause large changes in organ position, and real-time detection/prediction of this kind of organ motion is an active research topic as it could improve radiation targeting accuracy, increasing the effectiveness of radiation treatment. Langen and Jones [2001] compile the results of several researches, revealing the large influence of respiration on organ motion (summarized in tables 4.1 to 4.3). Deep breathing tends to significantly increase the motion amplitude. The studies seem to indicate that the dominant direction of motion is along the superior - inferior axis (SI), although smaller displacements (in the order of a few mm) are also observed in the anterior-posterior (AP) and lateral directions.

<table>
<thead>
<tr>
<th>Cited research</th>
<th># of patients</th>
<th>Avg ± SD</th>
<th>Range</th>
</tr>
</thead>
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<tr>
<td>Weiss et al. [1972]:</td>
<td>25</td>
<td>11 ± 3</td>
<td></td>
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<tr>
<td>Harauz and Bronskill [1979]:</td>
<td>51</td>
<td>14</td>
<td></td>
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<td>Suramo et al. [1984]:</td>
<td>50</td>
<td>25</td>
<td>10 - 40</td>
</tr>
<tr>
<td>Davies et al. [1994]:</td>
<td>9</td>
<td>10 ± 8</td>
<td>5 - 17</td>
</tr>
<tr>
<td>Balter et al. [1996]:</td>
<td>9</td>
<td>17</td>
<td></td>
</tr>
<tr>
<td>Shimizu et al. [1999]</td>
<td>1</td>
<td>21</td>
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Table 4.1: Summary of liver motion studies, adapted from Langen and Jones [2001]. Values give the peak-to-through motion in millimeters as observed during normal breathing in supine position.

Tang et al. [2004] describe a data gathering procedure recording the motion of external skin markers and internal liver fiducials from several swines, with the purpose of developing a correlation model between skin movement and internal organ motion. Frequency analysis indicates that internal and external fiducial movement is in phase with the largest frequency component between 0.18 and 0.21 hertz (equivalent to 11 to 12.5 breaths per minute).
Figure 4.4: Central arterial waveform (lower panel) and peripheral waveform (upper panel) in a young (right) and an elderly (left) subject. Adapted from Wilson et al. [2001].

Figure 4.5: Example of an actual ECG trace (Lead I) measured of a healthy young subject showing respiratory sinus arrhythmia. The R-R interval during maximum exhalation ($\Delta t_2 = 0.842$ sec.) is 13% higher than during inhalation ($\Delta t_1 = 0.7405$ sec.). (Note: The ECG signal is slightly distorted due to a presence of a 50 Hz noise and the use of a digital walking average filter used to repress this noise.)

Hostettler et al. [2008] describe the development of a real time simulation tool capable of predicting abdominal organ positions induced by free breathing, such that this tool can be used to increase radiation treatment accuracy. It is shown that the position of the main abdominal organs correlates
Table 4.2: Summary of diaphragm motion studies, adapted from [Langen and Jones 2001]. Values give the peak-to-through motion in millimeters as observed during normal breathing in supine position. [Korin et al. 1992] recognizes that the dominant direction of motion for the upper abdominal organs is along the Superior - Inferior axis.

<table>
<thead>
<tr>
<th>Cited research</th>
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<th>Avg ± SD</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Wade [1954]</td>
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<td>17 ± 3</td>
<td></td>
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<tr>
<td>Weiss et al. [1972]</td>
<td>30</td>
<td>13 ± 5</td>
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<tr>
<td>Korin et al. [1992]</td>
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<td>13</td>
<td>7 - 28</td>
</tr>
<tr>
<td>Davies et al. [1994]</td>
<td>9</td>
<td>12 ± 7</td>
<td>5 - 17</td>
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<tr>
<td>Hanley et al. [1999]</td>
<td>5</td>
<td>26.4</td>
<td>18.8 - 38.2</td>
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</table>

Table 4.3: Summary of kidney motion studies, adapted from [Langen and Jones 2001]. Values give the peak-to-through motion in millimeters as observed during normal breathing in supine position.

<table>
<thead>
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<th>Cited research</th>
<th># of patients</th>
<th>Avg ± SD</th>
<th>Range</th>
</tr>
</thead>
<tbody>
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<td>Suramo et al. [1984]</td>
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<td>10 - 40</td>
</tr>
<tr>
<td>Moerland and van den Bergh [1994]</td>
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<td>19</td>
<td>2 - 35</td>
</tr>
<tr>
<td>Davies et al. [1994]</td>
<td>8</td>
<td>11 ± 5</td>
<td>5 - 16</td>
</tr>
<tr>
<td>Balter et al. [1996]</td>
<td>18</td>
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</table>

with the patient skin motion and that real-time tracking of skin motion can be used to estimate organ position. Motion is estimated by modeling a thoracic and an (incompressible) abdominal envelope, both sharing the diaphragm as boundary. It is assumed that respiration will cause a volume increase of the thoracic envelope, and the presence of the thoracic cage forces expansion through the diaphragm. The SI motion of the diaphragm forces the (incompressible) abdominal envelope to expand in AP direction in order to retain its volume. Measuring the change in skin position reveals the displacement volume and the displacement field of the diaphragm. Refining diaphragm displacement in incompressible sagittal (e.g. dividing between left and right sections) slices results in local displacement estimates of the abdominal organs. Evaluation of this tool showed that this method is able to predict real-time abdominal organ position with a prediction accuracy within 2 - 3 mm when compared with CT images.

Chowdhury and Majumder [1981] performed a spectral analysis of the respiratory sounds. Analysis indicates a more or less maximum amplitude of about 250 Hz for subjects without pathological lung history, with rapid decrease in amplitude as the frequency increases and approaches 1000 Hz. In the case of the tubercular lung (fibrotic type), a significant frequency decrease of the maximum amplitude peak accompanied by the presence of higher frequency components, can be observed. Similar conclusions are made by [Dellinger et al. 2008] and [Wang et al. 2009].

Dellinger et al. [2008] describes the evaluation of a new computer-assisted lung sound imaging system, consisting out of the real-time processing of the vibration recorded by 2 arrays of piezoelectric sensors. These sensors are positioned on the back of the subject, such that they record the vibrations generated by lung activity. An algorithm is used to process the signals in the 150 - 250 Hz range.

Wang et al. [2009] use the same system in a pilot study was to determine whether patients with acute dyspnea due to Obstructive Airway Disease (OAD) had distinguishing features when studied with a computerized acoustic-based imaging technique. They have concluded that it was indeed possible to distinguish patients whose dyspnea was due to OAD from patients whose dyspnea was not caused by OAD. Differentiation of OAD was based on three distinct features: heterogeneity of the sound distribution in OAD, distribution of vibration energy between inspiration and exhalation, and the presence of a significantly longer exhalation phase in OAD, leading to a change in the Inspiration/Exhalation time ratio.
Table 4.4: Summary of pancreas motion studies, adapted from [Langen and Jones 2001].
Values give the peak-to-through motion in millimeters as observed during normal breathing in supine position.

<table>
<thead>
<tr>
<th>Cited research</th>
<th># of patients</th>
<th>Avg ± SD</th>
<th>Range</th>
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<tr>
<td>Suramo et al. [1984]</td>
<td>50</td>
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<td>10 - 30</td>
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<tr>
<td>Bryan et al. [1984]</td>
<td>36</td>
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Gastrointestinal motion

Activity of the gastrointestinal (GI) system is well-known vibration source, most commonly experienced by so called bowel sounds. Listening for these vibrations, audible in the form of bowel sounds (auscultation), is an essential part of the physical exam of the abdomen. During auscultation, the diaphragm of the stethoscope is pressed against the abdominal wall, and vibrations of the abdominal wall will be made audible for the physician. Absence of bowel sounds can indicate intra-abdominal infections, while the presence of characteristic high-pitched bowel sounds can be an early indication of small bowel obstruction. [Ferguson 1990].

Politzer et al. [1976] reveal the influence of viscus content and the viscus location on the genesis of bowels sounds. Applying perfusion revealed that the stomach is the most active site in production of bowel sounds, followed by the colon and the small bowel. Two types of sounds where distinguished: type NS which show sudden peaks of amplitude, but a relative low average amplitude over the defined time period. type NE sounds are characterized by larger average amplitudes, and shows a limited, high amplitude spectrum around 100 Hz and more dispersed frequency centered around 300 Hz.

Craine et al. [2002] performed 2d mapping of the abdomen surface and observed a spatial distribution of spectral power. Most sound originates from the right lower quadrant, possible generated by the ileocecal valve (the sphincter muscle at the junction of the small intestine and the large intestine). The second location is the stomach region, which shows a predominance of spectral content between 184-248 Hz in the stomach area, but also show increase of spectral power in the 350 - 410 Hz and 450 - 550 Hz frequencies. Healthy subject appear to have a 32.8% ± 20.0 of total sound power. IBS and NUD patients generally show a similar amplitude ath this region in as the other two frequency regions.

Tomomasa et al. [1999] recorded the abdomen sounds with a microphone positioned below the umbilicus of infants. the magnitude of the recorded sound, the Sound Index (SI) is correlated to the motility of the gastrointestinal track. An increase to normal values of SI is observed after successful treatment of hypotrophic pyloric stenosis (HPS, a motility-related disorder that is common in children), indicating a normalization of bowel motility.

Yoshino et al. [1990] show that bowel sound analysis can be used as an assessment of the severity of mechanical obstruction of the intestine. Discriminating into three distinct types of bowel sounds, distinguished on their respective frequency pattern, it is claimed that the method provides a very objective assessment of the severity of the obstruction, and could be helpful in the determining the treatment for each patient. During the analysis, ‘unique’ periodic frequency bands for cardiac sounds (20 - 200 Hz) and respiratory sounds (200 - 400 Hz) are identified and removed before further analysis. Normal bowel sounds are characterized as having a lower frequency of 198 ± 11 Hz, a peak frequency (the most audible sound) of 316 ± 27 Hz, and an upper frequency of 500 ± 35 Hz. Depending on the severity of the obstruction, peak frequency increases from 316 Hz to 612 Hz; the first being a healthy subject, the latter correlating with subjects with a 100 % operation rate.

Campbell et al. [1989] presented an alternative method, based on so called Surface Vibration Analysis (SVA), which relates gastrointestinal activity to SVA energy values. Vibrations [40 - 10000 Hz range] of the abdominal wall are registered by an accelerometer, whose signal content
is integrated over time, resulting in the so called SVA energy values. Measurements reveal a
correlation between propulsive gastrointestinal motor function and an increase of vibration energy.

4.3 Measurement setup and protocol

This section describes the final measurement setup used during this research. It mentions the used
subject position, positioning and use of sensors and summarized the measurement settings used
during this research.

It should be noted that most measurements (except for the measurements performed on AAA
patients described in section 6.6), are performed on a single young and healthy subject and as such
only contain inter-patient variability. The measurements described in this particular section have
been acquired during several measurement sessions and represent subject condition on different
time instances along a period of a couple of weeks.

4.3.1 Measurement setup

Subject position

Subjects position is quite similar to the position required when applying a physical examination
[Ferguson [1990]]. The subject is positioned supine (with face up) with his head supported by a
small pillow. The subjects arms are placed at the sides and not folded behind the head, as this
tenses the abdominal wall. Although [Ferguson [1990]] advise to support the hand and knees
with small pillows or folded sheets in order to comfort the subject and to relax the abdominal
musculature, it was decided not to use these additional supports. Consultation with the subject
revealed no particular discomfort and the subject abdominal muscles turned out to be relaxed
without the need of supporting knees and hands.

Although the abdominal aorta is located quite deep within the abdomen (the abdominal aorta
and inferior vena cava lie just above the spinal column), the spinal column restricts direct access
to the abdominal aorta. Due to this restriction on its back, pulsation of the AA is mostly directed
towards the abdominal wall, and during palpation the pulsation of the AA is evaluated by deep
palpation of the upper abdomen. [Fink et al. [2000]] describes the procedure to detect AAA as
‘placing both hand on the abdomen with palms down and an index finger on either side of the
pulsating area to measure aortic width’.

LDV position

The LDV is only able to detect motion perpendicular to the beam direction. In order to minimize
the signal variation caused by the curvature of the abdomen, it is necessary to position the detector
as far as possible from the abdomen wall. Especially during inter-patient measurements, it is
tried to keep the the relative position between detector and measurement surface constant. As
mentioned above, pulsation of the abdominal aorta is usually evaluated at the upper abdomen, so
the LDV is directed at the upper and middle part of the abdomen, with particular emphasis on
the surface area above the umbilicus.

It is possible to improve signal quality by positioning the object in a maximum of laser intensity.
With the laser used, these so-called optimal stand-off distances are at 99 mm, 303 mm, 507 mm,
711 mm, etc., measured from the front panel of the scanning head [Polytec Manual]. In normal
operation the intensity of the laser beam should not deteriorate to much, but efforts are taken to
obtain distances near the optimal stand-off distances.
Two different LDV positions were used during the research. During the measurements described in this chapter (section 4.4) and the measurements performed at AAA patients (section 6.6), the LDV was placed on a tripod and positioned such that the LDV is placed above the umbilicus as much as possible. All other data were measured using a custom made frame, whereby the LDV is positioned above the umbilicus.

![Figure 4.6: The proposed LDV setup. An ECG measures the heart rate and triggers the LDV controller to measure a new gridpoint after each heartbeat. The scanning head of the LDV is positioned above the subject abdomen, such that the object is in a maximum of laser intensity. The output of controller processes is stored on the PC.](image)

**ECG electrode positions**

Although the ECG is only used to trigger the scan, its hardware requires the positioning of 3 cardiogram paths. Following the recommendation of Dr. ir. C.A. Swenne (Associate Professor Cardiovascular Physiology Cardiology Department LUMC), a modified Einthoven’s triangle is used. A detailed description of this technique can be found in appendix [A.1](#). Besides the output of the three ECG leads (which will not be recorded during the measurements), the recorder also generates a 5V signal during the QRS complex of the cardiac cycle. This block signal is used as a trigger to shift the scan to a new scan point.

### 4.3.2 Finometer

During the measurements on AAA patients (described in section [6.6](#)), a finometer will be included in the setup (figure [F.7](#)). Using this addition is will be possible to obtain an input signal representing the excitation of the abdominal aorta, making it possible to define a frequency response function between blood pressure waveform and skin vibration. A pressure cuff, containing a build-in infrared photo-plethysmograph, is placed on the middle finger index (figure [4.7(b)](#)). The arterial pressure
in the finger is measured by means of the volume-clamp method [Bogert and Van Lieshout [2005]].

The finometer registers the brachial blood-pressure waveform and the results are logged on the LDV as a voltage signal.

Figure 4.7: The measurement setup used during the clinical test done at the LUMC. Figure 4.7(a) shows the finometer used to log blood pressure during the measurements. Figure 4.7(b) shows a detail of the handcuff used to detect blood pressure. A pressure cuff is placed on the middle finger index and inflated/deflated such that the bloodflow detected by the build-in infrared photo-plethysmograph is kept constant (Volume-clamp method). Figure 4.7(c) shows the complete setup of LDV components, ECG monitor and finometer.

4.3.3 Measurement protocol and data processing

Initially it was the intention to include measurements on a bare skin and compare performance with measurements obtained by using retro-reflective material. The measurement performed with retro-reflective material could act as a benchmark and would be used to assess the additional error caused by the use of bare skin. However, preliminary measurements showed that using reflective material resulted in superior results, resulting in the decision to postpone further measurements applied on bare skin.

As such, a grid of retro-reflective stickers was placed on the subjects abdomen, and gridpoints where defined in the center of these stickers (one gridpoint per sticker). Over the course of these experiments, the amount of stickers increased, refining the distribution and also extending the grid to the lower part of the thorax.

The patient was asked to avoid movement during each scan; he was also asked to refrain from talking during scanning in order to reduce noise generation. Due to the application of the reflective stickers, the subject was also required to remain still between the individual scans, such that the gridpoints (defined within the LDV software) would remain centered on the stickers. Stickers where usually spaced 15 to 20 mm. apart.

LDV output was logged in two diverent ways. Due to restrictions caused by software licensing, the earlier measurements had to be logged as a frequency spectrum. The LDV applies the Fast Fourier Transform on the recorded velocity or displacement profile, applies complex averaging the results, and finally logs the averaged frequency spectra. A MATLAB algorithm is used to recreate a time representation. Later measurements where directly logged as an amplitude-time signal, such that a more ‘raw’ signal was obtained.

Signal bandwidth was limited to the frequency content associated with cardiac activity and was set on a 0 - 100 Hz band. A low-pass filter with a cut-off frequency of 50 Hz was included in an effort to exclude non-cardiac noise from the signal. Literature (section 4.2) showed that cardiac vibration
and sounds are generally associated with the 0 - 50 Hz region. Respiration and Gastro-Intestinal vibrations sounds are usually associated with frequencies above 100 - 150 Hz.

In order to keep acquisition time at tolerable lengths, frequency resolution was usually set on 100 FFT lines, obtained on 1000 ms long sample periods. Frequency spectra where usually calculated after complex averaging of four separate acquisitions. The influence of drop-outs (sudden short loss of signal, visible as large spikes in the signal) was minimized by applying Speckle Tracking in the setting FAST.

4.4 Experimental Results

4.4.1 Influence of skin treatment on abdominal scans

The first acquisition was the first attempt too register abdomen motion. These initial tests only registered the first 50 FFT lines and only measured the frequency content of the first 500 ms (table 4.5). Measurements where performed using both bare skin and skin with retro reflectors applied on the abdominal surface. Triggering on the heartbeat seems to result in quite repeatable measurements. It was observed that the attempts to measure on bare skin causes ‘spikes’ to appear in the signal, distorting the FFT calculated by the LDV. Using retro reflective stickers result in a much clearer signal. Figures 4.8 and 4.9 illustrate the difference between obtained signal quality. Both signals are obtained after applying the inverse Fast Fourier Transformation on the frequency spectra logged by the LDV. Figure 4.8 is obtained by scanning the bare abdominal surface, and it can be observed that the signal is quite ‘spiky’. In contrast, the signals shown in 4.9 appear a lot smoother. It should be noted that the spatial distribution of the grid points between both acquisitions is not similar, the figures only illustrate the difference in signal quality and should their signal content should not be compared.

<table>
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<td>Samp. Time(ms)</td>
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<td>0 Hz - 100 Hz</td>
</tr>
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<td>-</td>
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<tr>
<td>Decoder:</td>
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<td>VD-06</td>
</tr>
<tr>
<td></td>
<td>50mm/s/V LP</td>
<td>50mm/s/V LP</td>
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</tbody>
</table>

Table 4.5: Settings used during the measurements in session 1. LP = Low Pass filter, VD = Velocity Decoder.
Figure 4.8: Several scanpoints of surface velocity obtained on bare skin. The signal is quite noisy in respect to the signals shown in figure 4.9, obtained with retro reflective stickers. It is worth to note that spatial distribution of the gridpoints used in this acquisition and the one shown in figure 4.9 are not similar. Comparison between both illustrates the difference in signal quality and should not be used to compare signal content.

Figure 4.9: Several scanpoints of surface velocity obtained with retro-reflective stickers applied on the abdominal surface. The signal is a lot less noisy than the signals shown in 4.8. Due to different spatial distribution of the gridpoints the comparison between both figures cannot be used to compare signal content.
4.4.2 Influence of breath holding techniques on measurements

In order to evaluate spatial distribution of the abdominal motion pattern, these measurements where performed with finer grid. Furthermore the grid is expanded towards the upper abdomen and the lower part of the thorax. In an effort to reduce the influence of possible noise, the frequency spectrum was logged after applying complex averaging of four separate recordings.

The sample period was increased to 100 FFT lines, effectively resulting in a sample time of 1000 ms. As an result complete cardiac cycles and part of the succeeding secondary cycle are recorded within each sample (table 4.5).

Besides a number of velocity measurement, a few attempts where made to obtain a displacement recording. It was observed that the displacement of the abdominal surface as highly influenced by respiration (as illustrated in figure 4.10 which compares the displacement measurement performed with normal breathing and when the subject was asked to hold his breath). Removing the 1 Hz frequency component out of the frequency spectrum before applying the iFFT during post processing did yield a more repetitive signal, showing more resemblance to the velocity recordings. However, the single recordings also showed that displacement measurements are vulnerable for signal drop-out, which up to now, has not yet been observed in the velocity recordings.

The influence of respiration was tested during a velocity measurements, concluding that holding breath did not alter the recording to a significant extend (figure 4.11). In accordance with this observation, it was concluded that it would indeed be possible to obtain sufficiently detailed measurements without the need of breath holding techniques during acquisition. The current long-time duration of the acquisition (up to 15 minutes per recording) was still deemed acceptable for use in the preliminary measurement, but continuous efforts were taken to reduce acquisition time.

<table>
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<td>LP &lt;50 Hz</td>
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<td>DD-200 640µm/V</td>
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<tr>
<td>Averaging:</td>
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</table>

Table 4.6: Settings used during the measurements in session 2. LP = Low Pass filter, VD = Velocity Decoder, DD = Displacement Decoder.
Figure 4.10: A comparison between the displacement of scanpoints measured with and without the subject holding its breath. The upper graphs (figures 4.10(a) and 4.10(b)) were both measured with normal breathing. During the acquisition of the lower graphs (figures 4.10(c) and 4.10(d)) the subject was asked to hold his breath. The upper graphs show a clear distortion caused by the slow respiratory rate, the lower graphs show a more repetitive signal. Both graphs are the averaged signals after 4 times complex averaging.
Figure 4.11: A comparison between the velocity of scanpoints measured with and without the subject holding its breath. The upper graphs (figures 4.11(a) and 4.11(b)) were both measured with normal breathing. During the acquisition of the lower graphs (figures 4.11(c) and 4.11(d)) the subject was asked to hold his breath. No significant difference is observed between the time signals of both situations. Both graphs are the averaged signals after 4 times complex averaging.
4.4.3 Influence of dropout and repeatability of measurements

A fine grid of retro reflective stickers is applied on the abdomen and the lower part of the thorax. In order to reduce the chance of a sticker moving out of the laser beam, sticker size was increased and more care was taken to ensure that the beam would hit the stickers in their center. This effort proved successful, the reflectors keep reflecting the beam and signal quality was a lot more consistent.

The previous experiences showed that displacement measurements where quite sensitive for drop-outs. During this session it was tried to reduce the amount of dropout by increasing the measurement range, with moderate success (settings summarized in table 4.7). Although dropouts were still present, the chance of it occurring did reduce. Measurements showed that displacement was dominantly influenced by respiration and (because cardiac cycle and respiration cycles do not synchronize) appeared ‘random-like’. Applying a bandpass filter to remove the frequencies below 0.5 Hz appears to remove the interference caused by respiration (illustrated by figure 4.12). Velocity measurements do not seem to be affected when holding breath or breathing normally.

<table>
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<tr>
<td>Samp. Time(ms)</td>
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<td>Filt set.:</td>
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<td>BP 0.5 - 50 Hz</td>
<td>BP 0.5 - 50 Hz</td>
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<td>Decoder:</td>
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<tr>
<td>Averaging:</td>
<td>4 * complex</td>
<td>4 * complex</td>
<td>4 * complex</td>
</tr>
</tbody>
</table>

Table 4.7: Settings used during the measurements in session 3. LP = Low Pass filter, BP = Band Pass filter. VD = Velocity Decoder, DD = Displacement Decoder.

Comparison between different velocity acquisitions reveals that scanpoint motion is quite repetitive: figure 4.13 compares the velocity profile between 4 separate acquisitions, revealing a similar signal content, especially over the first 0.5 seconds of the signal.
Figure 4.12: A comparison between scanpoints with dropout occurring (figures 4.12(a) and 4.12(b)) and scanpoints where no dropout is observed (figures 4.12(c) and 4.12(d)) during the displacement measurement of test 6. Dropout result in a sudden increase to the maximum amplitude of the range. Quite some repetition can be observed in the displacement signal if no drop-out occurs and signal content shows more resemblance with velocity profiles.
Comparison of time signals between scans.

Figure 4.13: A comparison between the averaged velocity - time signals of scanpoint 20 acquired during scans 1 - 4. It can be observed that especially the first 0.5 seconds are quite repeatable.
4.5 Conclusions

Literature revealed that it is possible to distinct two kinds of motion within the abdomen (and part of thoracic cavity). The low frequency organ motion can be defined as the motion of the viscera, most dominantly caused by respiration (e.g. the movement of the diaphragm). This motion is primarily along the superior - inferior axis and causes motion in the order of 10 to 20 mm. Motion in lateral and anterior - posterior directions are significantly smaller, about 1 or 2 mm, in contrast to the motion of the viscera, skin motion is primarily in the anterior - posterior direction: motion of the diaphragm causes a deformation of the ‘incompressible’ abdominal cavity (Hostettler et al. [2008]). Due to the stiff boundaries imposed by the pelvis and the muscular vertebral column, the easiest way for the abdominal cavity is to ‘bulge’ out towards the abdominal wall. Organ motion appears to correlate well with skin motion, to such an extend that its even possible to use skin motion to predict organ motion. The frequency of this cyclic motion is well below 1 Hz, in the order of 0.1 to 0.2 Hz, but is associated with large amplitudes.

Skin motion is also present in the form of sound, vibrations in the audible range. They are generally believed to be caused by the interactions of structures with fluids and the occurrence of turbulence. Sounds can be generated by the cardiac cycle (usually associated with sound / vibrations between 0.5 and 50 Hz), respiration (audible range, 150 Hz and higher frequencies) and gastrointestinal sounds (classified as peaking in amplitude above 100 Hz).

It can be concluded that cardiac induced vibration should be dominant in the 0.5 Hz till 50 Hz range, and it should be possible to reduce the influence of respiration and gastrointestinal activity by applying a bandpass filter on this range. However, respiration will also cause a displacement of the viscera (abdominal organs) in the order of several cm, and a relative displacement between the viscera and the scanned skin surface can be expected.

Based on the performed measurements it can be concluded that it is indeed possible to detect a repeatable motion of the abdominal surface, and it appears that the cardiac cycle plays a dominant role in generating this motion. In contrast to the initial expectations, beam reflection at bare skin introduces a problematic amount of noise, which possibly enhanced by the application of the FFT transform, reduces the applicability of the signal. Applying retro reflecting stickers on the abdomen surface seems to solve this problem.

Measuring the displacement of the abdomen surface proofed to be much more difficult than velocity measurements. Displacement measurements are a lot more sensitive for dropouts which, in combination of the use of complex averaging over multiple signals, results in large distortions in the recorded frequency spectra. Although the LDV software can be set up to discard faulty acquisitions, experience showed that this tends to result in excessive long scanning periods.

Respiration appears to be dominantly present in the displacement signal, but filtering signal content below 0.5 Hz seems to remove most of this influence, yielding results more comparable with the velocity measurements. It is expected that respiration also influences surface velocity, but due to the relative low velocities associated with respiration and the tendency of the FFT - iFFT processing algorithm to leave DC offset out of the signal, it is not that clearly visible within the recorded signals. During later experiments, involving longer acquisition times and the recording of actual velocity - time profiles, it is revealed that a low frequency system (most likely respiration) is present within the signal.

Due to limits imposed by the available software licenses, it was only possible to log frequency spectra. This restricted the acquisition to a sample rate of 200 Hz (100 separate FFT lines in the range 0 - 100 Hz) during each measurement. By means of the inverse Fast Fourier Transform it was possible to reconstruct a time signal, but its usefulness is affected by the restrictions imposed by the FFT. Because FFT assumes stationary harmonics, it is not capable of detecting fast transient signals or non-stationary signals, limiting post-processing possibilities.
Evaluation of dataprocessing techniques

5.1 Introduction

During the previous experiments (described in chapter 4) the abilities of the LDV setup have been evaluated. The results showed that it was possible to obtain repeatable measurements and it was possible to retrieve the velocity and displacement profiles by inverse Fast Fourier Transform of the logged frequency spectra. Furthermore, the measurements revealed that the cardiac cycle plays a dominant role in motion generation of the surface, equivalent to the results obtained during the very first attempts to detect surface motion at a human wrist (chapter 3). This chapter focuses on available data analysis techniques which could be used during data processing. Along the course of the research project, a multitude of techniques have been tried, and this chapter tries to summarize their strong and weak points and the experiences obtained during their application.

It is possible to distinguish two different fields of operation along the analysis techniques. Some techniques are focused on the time domain, other techniques focus on the spatial distribution of the signals, and others are a hybrid of both fields. However, a trend of increased decomposition can be observed along both of the fields: a decomposition from time towards frequency components and the trend of decomposition from spatial location towards spatial wave frequency components (identifying the shape of the accounted vibration patterns along the abdomen surface). Figure 5.1 shows a schematic summarizing the techniques and indicating before mentioned trends.

5.2 Signal origin

Two different data acquisition methods have been used during the project. During the initial phase of the research, described in section 4.4 of the previous chapter, the vibration of the abdomen surface was logged using (complex averaged) FFT spectra. In a later phase, the abdomen vibration was logged using velocity - time representations. Both methods seem to have their distinct advantages and disadvantages, and their features have some impact on the results of further data processing:

- The LDV software does not log the 0 Hz DC component within its FFT spectrum. This tends to reduce the influence of respiration when compared with the amplitude - time logged data.
• The LDV software can automatically calculate the complex average before logging the signal as an FFT, reducing the influence of noise, but at the expense that information regarding the standard deviation of the complex average is lost.
Due to the application of a rectangular window during the acquisition, the logged spectrum contains influence of spectral leakage. This can be (partially) resolved by reconstructing the velocity-time signal by means of the inverse Fourier Transformation, applying a window function on the sample and recalculating the FFT.

The application of complex averaging causes the LDV to measure each gridpoint several times before proceeding to the next gridpoint. This yields an increase of acquisition time per scanpoint, taking the LDV quite a long period to complete a single scan of the entire abdomen surface. Due to this long scantime, the approach of approximating a parallel scan by a sequential scan becomes less accurate. The system generalizes the motion occurring over a long period and assumes it represents the motion for a brief instant. However, over the course of this longer period, a low frequency trend could be present and as a result, mistakenly be identified as a fast occurring process.

5.3 Signal Magnitude analysis

Signal Magnitude analysis is a common method based on identifying the overall magnitude (r.m.s or peak) of a signal, for instance mechanical vibration or sound levels that pass a certain maximum level. Thresholds are defined on the basis of mechanical models or based on experience. This approach assumes that relevant changes in a process are sufficiently contained within the amplitude of the system's output. Analysis of magnitude level, or the statistics contained within signal magnitude (probability densities, mean values, variance, skewness or analysis of extreme values) in turn reveal information regarding the changing system. This approach is for instance used for bearing fault detection or noise level evaluation.

As mentioned in section 4.2 of the last chapter, the internal abdomen is not a static or quasi static system. Organs move under influence of respiration, inputs like respiration and the cardiac cycle change in cycle rate. Blood pressure waveforms change in shape and the pulse pressure within the arteries fluctuates. As such, the input, or excitation sources for the abdomen vibration vary considerably. Focusing purely on threshold or variation in magnitude contains the risk of identifying changes in input signal, instead of changes of the transmission path. Suitable analysis methods should at least be able to track changes in the input and it is questionable if signal magnitude analysis is sufficient on its own. One particular approach, although not worked out sufficiently within this research, could be the identification of unusual spatial distributions of energy, such as used for lung efficiency analysis [Wang et al. 2009] or bowel sounds [Craie et al. 2002].

5.4 Time Domain Analysis

All Time Domain analysis methods which are based on direct interpretation of the velocity profiles suffer from the influence of respiration. As will be treated in the next chapter, section 6.3, respiration will be visible in the form of a low frequency offset above the cardiac cycle induced vibration, and as such distort direct interpretation methods. Because respiration is independent of the cardiac cycle (the contrary is not true, respiration can indeed influence the heart rate), averaging of the time signals tends to average all the non-cardiac related signals out. This can reduce the influence of respiration to some extend. Some of the analysis methods discussed within this section exploit this feature.
5.4.1 Instantaneous Deflection Surface

A multitude of different time domain analysis methods can be applied, one of the most natural would be the definition of an instantaneous deflection surface, where the obtained velocity profiles are used to define the motion of a surface. An important benefit of this approach is the clear visualization of the spatial distribution of the measured velocity profiles. It does however require a certain degree of synchronization between the velocity profiles in order to produce usable results: due to variability in the heartbeat rate (up to 15% in healthy, young individuals, as previously identified in section 4.2.1), the synchronous nature will be mostly limited to the first systole observed in the measurement, as the time between the trigger and the second systole will vary. (It is however possible to scale the velocity signal according with the beat to beat period, with methods used in ECG analysis. These methods have not been used during this research.)

![Velocity profiles for each meas. gridpoint](image)

**Figure 5.2:** Instantaneous Deflection Surface, plotted for 3 different time instances. Subfigure 5.2(a) shows the (usable) velocity profiles measured at the gridpoints and identifies time instants a, b & c. Subfigures 5.2(b), 5.2(c) and 5.2(d) show the accompanying Instantaneous Deflection Surfaces at time instants a, b and c respectively. The surfaces are generated using triangle based cubic interpolation between the gridpoints.

Figure 5.2 illustrates examples of an Instantaneous Deflection Surface. The IDS has been plotted for three different time instances occurring during the cardiac sine. This particular example shows the IDS of a single measurement scan, and as such, still contains the influence of respiration. This causes a certain degree of asynchronous motion between the different measured velocity profiles.
This approach specifically focuses on instantaneous spatial distribution, and temporal information regarding the velocity profiles along time domain is not evident. This drawback can be partially solved by animating the IDS along the time domain, but this visualization technique will remain limited in its ability to show the velocity waveform along the time domain.
5.4.2 Velocity distribution

While the previous analysis method almost primarily focused on the spatial distribution, it is also possible to emphasize on the temporal information instead of spatial distribution. This can be achieved by plotting all the velocity profiles, and arranging the profiles according to the locations of the gridpoints on the abdomen surface. This approach is illustrated in figures 5.3 and 5.4, in which the former shows all of the obtained velocity profiles for each scanpoint (this dataset contains seven separate scans, each scan shown as an separate velocity profile) and the latter shows the average velocity profile (the average is calculated for the seven available profiles within this dataset). Figure 5.5(b) shows a photograph of the real gridpoints and their position on the abdomen. It should be noted that some gridpoints have been deleted from the analysis, because their measurements were considered to be too noisy to be processed.

Although this method emphasizes the temporal information, it is also quite helpful in detecting spatial trends. For instance, the dataset displayed in figures 5.3 and 5.4 reveal the appearance of the so called cardiac sine at the line left of the umbilicus, an area where the Abdominal Aorta normally should be located. Further away from this position, the velocity profiles become significantly more complex, and it becomes harder to identify a cardiac sine like waveform.
Figure 5.3: Individual velocity profiles for each gridpoint, ordered in respect to the gridpoint positions on the abdomen surface. Each individual subplot shows all of the measured velocity profiles available for each gridpoint (this particular dataset is obtained from a healthy individual and contains 7 individual scans). Some spread can be observed, most likely under the influence of respiration.
Figure 5.4: Individual averaged velocity profiles for each gridpoint, again ordered in respect to the gridpoints positions. The averaged velocity profiles are obtained by averaging the datasets shown in figure 5.3. Averaging reduces the influence of non-periodic signal components (like respiration) and will emphasize the periodic parts.
5.4.3 **Velocity Line averaging**

Inspection of the individual velocity profiles (for example as displayed in figures 5.3 and 5.4) reveals that the velocity profile of the skin surface is dependent on location. It is also observed that the time instance of the ‘cardiac sine’ appears to be dependent on the height on the abdomen, gridpoints positioned lower on the abdomen show a slightly larger delay (This phenomenon is treated more thoroughly in section 6.2). Unfortunately, the previous approach of plotting the individual velocity profiles for each scanpoint separately, does not result in a clear visualization on this tendency.

An alternative method is to apply Line Averaging on the velocity profiles: by averaging all the velocity profiles of the gridpoints with (roughly) the same y-coordinate (as defined in figure 5.5(a)), the individual velocity profiles shown in figures 5.3 and 5.4 can be compiled in seven line averaged velocity profiles. Figures 5.6 & 5.7 show the resulting average velocity profiles, the 95% two-sided confidence intervals (using eq. 5.1) and the standard deviations around the estimated average.

Students t-distribution is used to calculate the confidence interval for the calculated mean of each dataset ([Dekking et al. 2004]):

\[ \bar{X}_n \pm t_{p/2, n-1} \frac{S_n}{\sqrt{n}} \]  

(5.1)

Where:

- \( \bar{X}_n \) = mean of the dataset
- \( t_{p/2, n-1} \) = the integral of Student’s probability density function, dependent on the required accuracy \( p \) and number of observations in the dataset \( n \)
- \( S_n \) = sample standard deviation
- \( n \) = number of observations in the dataset

This approach has got some benefits over the previous analysis of individual velocity profiles. Due the compilation of individual profiles into of a single averaged profile, it simplified the analysis and makes it easier to compare between datasets. It also has got the additional benefit that it calculates the average over a larger dataset (it averages over multiple gridpoints, each containing the profiles of a multitude of scans), thereby reducing influence of respiration by averaged it out of the results. However, this approach also holds some serious disadvantages. It loses a lot of information regarding the spatial distribution along the x-direction (width of the abdomen), which as figure 5.4 previously illustrated, contains relevant information. Furthermore, in order to analyze the propagation of the wave velocity of the blood pressure waveform, it would be better easier to restrict the analysis to gridpoints lying above the abdominal aorta. This approach is used in section 6.2 to estimate the wave velocity of this phenomenon.
Figure 5.5: Subfigure 5.5(a) shows the line definition used during Line averaging. The velocity profiles of gridpoints (numbered) with roughly the same y-position are averaged to a single profile. Gridpoints with a lot of noise in their velocity profiles are omitted from analysis (crossed out points). Subfigure 5.5(b) shows the position of the reflective stickers on the subjects abdomen.
Figure 5.6: Averaged velocity profiles of a healthy subject for the vertical gridlines L1 - L3. The plots show the mean velocity profile, the t-distribution estimated confidence intervals and the standard deviation. The velocity profile is obtained after averaging the velocity profiles of all gridpoints associated with the gridlines for this particular dataset. The relative large standard deviation can most likely be attributed to respiration and to some extend spatial variation, especially considering the relative tight confidence intervals around the estimated mean velocity profiles.
Figure 5.7: Averaged velocity profiles, the t-distribution estimated confidence intervals and the standard deviation of a healthy subject for the vertical gridlines L4 - L6 (subfigure 5.7(a)) and for line L7 (subfigure 5.7(b))
5.5 Frequency - Time Domain analysis

Frequency - Time domain analysis methods are required in order to be able to analyze the temporal variation of frequency content. This section will discuss two methods which can be used to obtain this information: the Short Time Fourier Transform (STFT) and the Continuous Wavelet Transform (CWT). Due to this advantages, only the CWT has been used as an analysis tool within this research, but the STFT is included in order to provide a more complete overview of available methods.

5.5.1 Short Time Fourier Analysis

The Short Fourier Transform (STFT) is obtained by applying the Fourier Transform on local sections of the signal and yields the frequency content of each evaluated section. This method is defined on a fixed resolution, and requires a trade off between time and frequency resolution. A detailed description of this technique is available in appendix D.2.5.

Due to the relative small sample times, especially when compared with the frequency band of interest (0 - 50 Hz), application of the STFT will be extremely limited. For example, splitting the acquired window lengths in distinct sections of 0.25 sec (roughly a quarter of a normal heart cycle) will result in a frequency resolution of at most 5 Hz.

In order to reduce spectral leakage, it is also required to define the convolution of a window function with the sample signal, before calculating the Fast Fourier Transform. Application of a window function will deteriorate the frequency resolution even further. Although this deterioration can be partially resolved by using overlapping window functions, the frequency resolution (especially at the lower frequencies) will remain impractically low.

5.5.2 Continuous Wavelet Transform

The Wavelet Transform (WT) is an alternative method specifically designed to analyze the non-steady frequency content within a signal. While the Fourier Transform decomposes the signal in a summation of stationary sinusoids of infinite duration, the WT decomposes a signal in wavelets, non-periodic, non-steady signals of finite duration. This makes the method more suitable for the evaluation of localized frequency content. Unlike the STFT, the WT uses a variable resolution: higher frequencies are better resolved in time, but will have lower frequency resolution. Lower frequencies are resolved at a finer frequency resolution, but at the cost of reduced time resolution [Polikar 2010]. A more detailed description of this technique is available in appendix D.3.1.

Figure 5.8 illustrate the use of the CWT on a velocity profile, and the influence of windowing on the resulting CWT plots.
Figure 5.8: Examples of the 'coif4' Continuous Wavelet Transforms, calculated from the windowed velocity profiles shown in subfigure 5.8(a): the blue profile corresponds with the original velocity profile (CWT in 5.8(b)), the green profile uses a Blackman-Harris window (CWT in 5.8(c)), the red profile uses a 10% Tukey window (CWT in 5.8(d)).
5.6 Frequency domain Analysis

There are several different analysis techniques available which decompose a signal into the frequency domain. Several of these techniques have been used over the course of this research project, and this section will try to summarize their application and assess their usability.

This section will discuss the use of spectral density analysis (Auto Power Spectra), the use of Frequency Response Functions (FRF) and coherence functions. The results of a frequency analysis can also be used to define Operational Deflection Shapes (ODS) for each frequency, which in turn can be decomposed in a 2D spatial Fourier Transform, indicating which wavelengths and which directions are dominant in the vibration patterns of the abdomen surface. Finally the section will discuss the use of Cepstrum analysis, which can be used to identify periodic repetitions of frequency content within signals.

5.6.1 Auto Power Spectrum

The frequency content of the signal is evaluated using the Auto Power Spectrum (APS), and reveals the power distribution along the frequency domain. The APS is defined as the complex conjugate of the (Fast) Fourier Transform $X_{FFT}$ (equation 5.2). A window function is applied prior to calculating the Fourier Transform in order to reduce spectral leakage. The frequency resolution is artificially increased by inserting trailing zero’s after the windowed sample period (zeropadding). This increases the amount of samples fed in the FFT algorithm, and reduces sample spacing in the frequency domain. However, it should be realized that reducing sample spacing could also increase the visibility of spectral leakage in the FFT, mistakenly interpreting the effects of spectral leakage as frequency content. Choosing a suitable window function will reduce this distortion.

$$APS_{xx} = X_{FFT}(f) \cdot X_{FFT}^*(f)$$

(5.2)

Using a similar approach as used in figure 5.3, a FFT for each scan available within a dataset is calculated for each individual gridpoint. The FFT’s itself are complex averaged into a single average FFT, which is than used to define an Auto Power Spectrum. The results shown in figure 5.9 are obtained after applying 75% zeropadding and using a 10% Tukey window function.

In order to visualize the large range of spectral power, the vibration velocity power is transformed into a logarithmic scale defined by relation 5.3, referenced to $v_0 = 10^{-9} m s^{-1}$, an internationally accepted reference value [Norton and Karczub] (although usually referred to as a baseline for FFT vibration amplitude levels, instead of APS vibration power levels).

$$L_v = 20 \log_{10} \frac{v}{v_0} dB$$

(5.3)
Figure 5.9: Individual AutoPower spectra for each gridpoint, ordered in respect to the gridpoints positions. The original dataset contained 7 individual abdomen scans, and FFT spectra have been calculated for each scan. The AutoPower spectra are calculated after applying complex averaging of the individual FFT spectra. The spectra are zoomed in on the bandwidth 0 - 25 Hz and is scaled on a dB scale with \( v_0 \) as reference value. A 10% Tukey (tapered cosine) window function was used while using 75% zeropadding prior to calculating the FFT spectra.
The Auto Power Spectra shown in figure 5.11 are obtained after calculating the Fast Fourier transform, but this time averaged over multiple scanpoints. The abdomen surface is sub-divided in 4 quadrants (left / right and upper / lower abdomen) and it is assumed that these areas, lying quite close to each other, should have similar frequency content. First an average FFT spectrum is defined over multiple acquisitions for each scanpoint. Afterwards an average quadrant spectrum is defined by averaging the spectra of the grid points associated with each quadrant (figure 5.10) and the Autopower spectrum is defined. Again a Tukey window (10%) function is used in order to suppress the influence of spectral leakage within the spectra and 75 % zero-padding% zero is applied in order to increase frequency resolution within the evaluated bandwidth of 0 - 25 Hz. A relative standard deviation is calculated in order to visualize the deviation in amplitude levels between the FFT’s.

![Diagram](image)

**Figure 5.10:** Quadrant averaging: first the average Fast Fourier Transform of the dataset is calculated for each gridpoint separately. After this first step a complex average FFT is calculated using the spectra of the gridpoints associated with each quadrant. This averaged FFT is used to calculate the Auto Power Spectrum.
Figure 5.11: Example of Quad averaged Autopower Spectra. The abdomen surface is divided in four quadrants and an average APS (blue) is calculated of the individual FFT spectra for each gridpoint associated with a quadrant. The associated relative standard deviation (grey) shows considerable increase after 10 to 12 Hz. A 10% Tukey (tapered cosine) window function was used while using 75% zeropadding prior to calculating the FFT spectra.
5.6.2 Frequency Response Function

Previous analysis methods focused on evaluation of the output signal itself (skin vibration velocity), which is caused by both the input signal but which is also attenuated along the transmission path between input and output. By measuring both the input (blood pressure waveform) and the output (skin velocity) signals, it is possible to define the Frequency Response Function (FRF), describing how the abdominal surface velocity reacts under influence of the input blood pressure waveform. To do so, it is necessary to define the frequency spectra of both the blood pressure (input) and velocity (output) signals.

In order to reduce spectral leakage and (thus) to be able to apply zero-padding in order to obtain a high frequency resolution on the lower frequency band, it will be necessary to apply a window function on both signals. This is similar to the approach described in the previous section (section 5.6.1) and is described in more detail in subsections D.4.1 and D.5 of the Appendix.

As noted in figure 5.12, the blood pressure $P(t)$ contains a large DC component, which would dominate the frequency spectrum and, due to spectral leakage, could contaminate the spectrum at other frequencies. In order to resolve this problem, further analysis will be performed on the derivative of the pressure signal $\frac{dP}{dt}$. Because the blood pressure at the beginning and ending of each sample will be approximately equal, the ‘Fundamental Theorem of Calculus’ (equation 5.4) states that the DC offset of its derivative will approximately zero.

$$\int_{P(0)}^{P(T)} \frac{dP}{dt} (t) \, dt = P(T) - P(0) \approx 0$$ (5.4)

![Figure 5.12: Fino signal P(t) and derivative dP(t)/dt](image)

Figure 5.12: Fino signal $P(t)$ and its derivative $\frac{dP}{dt}$. $P(t)$ show a clear DC offset, its derivative shows a DC offset of nearly zero, consistent with the ‘Fundamental Theorem of Calculus’.

Figure 5.13 shows an example of the FRF and the coherence functions of an AAA patient prior to EVAR. Both functions have been plotted for each of the gridpoints after excluding several
particular noisy signals (corresponding to scanpoints where the reflector moved out of the laser beam). The FRF for each gridpoint is obtained after averaging Autopower spectra and Crosspower spectra of each acquired datasets, as defined in equation D.29 [appendix D.4.2]. It can be observed that the FRF’s appear quite similar. However, the coherence function is fluctuating, indicating that the assumed input signal is not the only source of the skin motion. However, it does show quite high coherence (between 0.8 till 0.95) between 5 and 9 Hz and Coherence above 2 Hz is still high for most of the involved of gridpoints. This could indicate that the output of most gridpoints can indeed be mostly attributed to the input bloodpressure waveform. It is interesting to note that the FRF appears to show resonance peaks at 5 and 9 Hz, but this is also accompanied with a sudden drop in coherence. Anti-resonance usually causes a drop in coherence due to lower signal to noise ratio at these ratios. At resonance, the sudden increase of amplitude of the (resonating) output also causes a drop in coherence, because the power of the output signal is suddenly much larger than the input signal.
Figure 5.13: FRF and Coherence functions of the pre-EVAR measurements performed an AAA patient prior to EVAR. The FRF’s and Coherence functions are calculated using the average APS’s and CFS’s over the available scans and plotted for each gridpoint, after removing noisy signals. The derivative of blood pressure $\frac{dP}{dt}(t)$ was used as input function, the velocity signal $v(t)$ was used as output. A Hann window function was used on $\frac{dP}{dt}(t)$, a Tukey (10%) tapered cosine window was used on $v(t)$. 
5.6.3 Operational Deflection Shape

An Operational Deflection Shape uses the frequency components of a FRF to define the surface deflection shape for individual frequencies. At resonance a ODS is essentially equivalent to a mode shape, because the surface should be vibrating in the mode shape of the resonance frequency. In order to evaluate the presence of the resonance peaks identified in a FRF (figure 5.13(a)), the Operational Displacement Shapes (ODS) for these specific frequencies can be calculated using relation 5.5. The resulting shapes are plotted in figure 5.14. If a frequency is indeed a resonance peak, the resulting ODS should have the appearance of a modeshape and show some sort of pattern.

\[
W_{n,f_i}(t) = |Y_n(f_i)| \sin \left( 2\pi f_i t + \arctan \left( \frac{\text{Im}(Y_n(f_i))}{\text{Re}(Y_n(f_i))} \right) \right)
\]  

(5.5)
Figure 5.14: The Operation Deflection Shapes (ODS) of the pre-EVAR measurements of patients 3 at 5, 7.5 and 9 Hz. Shapes 5.14(a), 5.14(d), 5.14(g) and 5.14(j) show the ODS of 5 Hz at $t = 0T, 0.25T, 0.5T$ and $0.75T$ respectively. Shapes 5.14(b), 5.14(e), 5.14(h) and 5.14(k) show the ODS of 7.5 Hz at $t = 0T, 0.25T, 0.5T$ and $0.75T$ respectively. Shapes 5.14(c), 5.14(f), 5.14(i) and 5.14(l) show the ODS of 9 Hz at $t = 0T, 0.25T, 0.5T$ and $0.75T$ respectively. Especially the ODS of 7.5 Hz appears to show a clear wave rolling along the vertical direction.
5.6.4 Cepstrum analysis

Cepstrum analysis is a relatively new analysis technique, based on power spectra. It can be seen as information about rate of change in the different spectrum bands. It can be used to identify the presence of a periodic structure within a power spectrum.

Two equivalent definitions for the Power cepstrum are available, both resulting in the same frequency spectral distribution, only differing by a scaling factor (equation 5.6) [Norton and Karczub]:

\[
C_{pxx}(\tau) = \mathcal{F}^{-1}\{\log_{10} G_{xx}(\omega)\}
\]

\[
C_{pxx}(\tau) = |\mathcal{F}\{\log_{10} G_{xx}(\omega)\}|^2
\]  

(5.6)

Where \(\mathcal{F}^{-1}\) represents the inverse Fourier transform of the term in brackets. The independent variable, \(\tau\), has the dimensions of time (similar to the time delay variable used in the definition of an auto-correlation function) and is referred to as the ‘quefrency’. \(G_{xx}(\omega)\) refers to the power spectral density (or Auto Power Spectrum) at frequency \(\omega\). The latter definition has the advantage that it is more efficient to calculate because it uses two forward Fourier transforms rather than an inverse Fourier transform.

Applications range from detecting periodic effects (such as harmonic patterns in machine vibration spectra), detecting and separating different sideband families in a spectrum or echo detection and removal. It can also be used to separate source and transmission path effects because it can distinguish both in readily identifiable quefrency peaks. The latter application is the result of the important property of the cepstrum domain in where the convolution of two signals can be expressed as the addition of their spectra (equation 5.7).

\[x_1 \ast x_2 = x'_1 + x'_2\]  

(5.7)

While the power spectrum does not contain phase information, the complex cepstrum (although being a real-valued function) does retain the phase information of the signal. As such, it is possible to define the complex cepstrum, discard unwanted quefrency components from the complex cepstrum, and transform back to the time domain, thereby obtaining the original time signal without the unwanted effects. This approach is used in echo removal and in the analysis of seismic signals in order to remove the seismic wave pulse caused by impulse response of the earth from the seismic signals [Norton and Karczub].

In order to illustrate the application of this technique, figure 5.15 shows the timesignal, the amplitude of its frequency spectra and the power cepstrum of an originally obtained signal and the same signal when artificially inserting an echo (adding the original signal with a time shift \(\tau = 1\) second to the signal). An harmonic can be observed at 513 samples, the time delay of the original signal, indicating a repetition of signal.
Figure 5.15: **Illustration of the Cepstrum analysis.** Subfigure 5.15(a) shows a velocity signal, its power cepstrum (subfigure 5.15(c)) and the magnitude of its frequency spectrum \(|FFT|\) (subfigure 5.15(e)). In order to illustrate the application of Cepstrums for echo-detection, an artificial echo is introduced by adding the timesignal 5.15(a) after a delay of \(\tau = 1\) second (\(v_2(t) = v(t) + v(t-\tau)\)). The power Cepstrum (subfigure 5.15(d)) indeed shows a Rahmonics at a Quefrency of 1 second (the red arrows show the first and second Rahmonics). The Frequency spectra (subfigures 5.15(e) & 5.15(f)) were not able to identify this echo.
5.6.5 2D Spatial Fourier Transform

Spectral analysis is usually associated with the transformation of a signal from the time to the frequency domain, and provides a decomposition of a time signal into frequency components given by sinusoids. However, it is also possible to decompose a surface wave into spatial wave components. Such an analysis renders information about the direction and composition of a surface wave, and can be a useful tool to analyses vibration patterns of a surface.

In order to obtain the FFT, a surface is defined by interpolated through the motion of the scan-points \( V_n(x_n, y_n) \), such that a motion profile \( V_\Omega \) is obtained for a grid \( \Omega \) of 50 x 50 gridpoints. Secondly a FFT \( Y_\Omega \) is calculated for each of the points of the grid \( \Omega \). Operational Deflection shapes are defined for the frequency \( \omega_i \) of interest, yielding a surface containing the amplitude (velocity) for each of the gridpoints for frequency \( \omega \). Finally, the 2D FFT is calculated over this surface and its magnitude is plotted in logarithmic scale. This approach is schematically summarized in figure 5.16.

**Figure 5.16:** Schematic of the 2D FT decomposition, summarizing the steps taken to obtain the 2D spatial Fourier Transform decomposition.

Figure 5.17 shows two examples of a 2D spatial FFT. The ODS of two distinct frequencies (4 and 8 Hz) have been decomposed in spatial wavelength components and the results are plotted in a logarithmic scale. the zero \( m^{-1} \) component is positioned in the center of the plot. The further away from the center, the higher the wave spatial frequency becomes.
Figure 5.17: Example of 2D Fourier transform. Subfigures 5.17(a) and 5.17(c) show the Operational Deflection shapes calculated for 4 and 8 Hz respectively. Subfigures 5.17(b) and 5.17(d) show the 2D Fourier transform their Operational Deflection Shapes.
5.7 Conclusions

This chapter described several analysis techniques which can be used to quantify aspects of the vibration pattern of the abdominal wall into measurable results. After experimenting with these techniques during the research, the following conclusions can be drawn:

Measurements of the abdomen surface yields information containing both a time distribution of by cardiac cycle induced vibrations, but also contains information regarding their spatial distribution. The obtained profiles are a result of the cardiac input, but also affected by the intermediate transmission path and are affected by other excitation sources. Depending on the type of analysis technique, certain aspects are highlighted. Time Domain analysis can place emphasis on both spatial or temporal signal magnitude, time and or frequency decompositions can be used to decompose the motion into simplified components. Its is even possible to decompose the vibration pattern first into frequency related Displacement modes and than analyze the spatial distribution of these patterns.

During the research it was found that the vibration pattern on the abdomen is quite complex. Most of the abdomen appears to show a periodic vibration, at least in synchronization with the cardiac cycle. Analysis of the abdomen surface above the Abdominal Aorta reveals a relative simple pattern, showing quite some similarity to the vibration patterns previously observed when measuring direct above the carotid artery. Further away, the signal still appears to be periodic with the cardiac cycle, but becomes significantly more complex. This complexity could be the result of transmission path effects, interference of other excitation sources or may be caused by misinterpretation of noise. It does however indicate that a proper analysis should be able to distinguish the location on the abdomen, and techniques like Quadrant averaging or Line-averaging which average the signal in time or frequency over some region simplifies the interpretation. This might hide some subtle features in the signal that could be related to local mechanical perturbations.

Analysis techniques like Frequency Response functions, Operation Deflection Surfaces and 2D Spatial Fourier Transform assume that the periodic input from the cardiac cycle excite the dynamics of the internal abdomen and that the observed vibration pattern is indeed caused by the mechanical properties of the system. Although FRF and Coherence plots indeed seem to suggest that dynamics like resonances occur, more thorough analysis is necessary to determine if this is the case. If it is indeed proven that standard conditions excite the internal abdomen into dynamic behavior, the use of Frequency Response Functions, Operational Deflection Shapes and 2D Spatial Surface transforms could distinguish relevant information.
Chapter 6

Influence of Physiological events on motion

6.1 Introduction

During the earlier experiments (described in chapter 4) the abilities of the LDV setup have been evaluated. The results showed that it was possible to obtain repeatable measurements and it was possible to retrieve the velocity and displacement profiles by inverse Fast Fourier Transform of the logged frequency spectra. Furthermore, the measurements revealed that the cardiac cycle plays a dominant role in motion generation of the surface, equivalent to the results obtained during the very first attempts to detect surface motion at a human wrist (chapter 3).

Chapter 5 described a multitude of analysis techniques and discussed their advantages and disadvantages on this particular application. It was recognized that abdomen surface motion can locally shows quite complex behavior, and it is required to take spatial distribution into account during analysis.

The next step is to use these techniques to evaluate the influence of ‘common’ physiological events on the resulting abdomen motion; the influence of sober and full stomach, tensioning of abdomen muscles and respiration cycle are measured and analyzed. Finally, in order to evaluate the influence of abdominal aorta pulsation on motion of the abdominal surface, measurements where performed on subjects diagnosed with Abdominal Aortic Aneurysm, a condition that should cause an increased pulsation of the abdominal aorta.

Two different data logging methods are used in this section; velocity profiles are logged in the form of a FFT spectra or in an amplitude-time representation, both yielding different results. As already explained prior, the FFT spectra are logged after applying complex averaging, which tends to average influence of respiration out and does not contain a 0 DC component. However, it is also observed that signal amplitude appears to be somewhat less and also, more striking, shows a different timing than observed in the amplitude-time representations. It is not known if this is an artifact caused by delay in the measurement system, caused by the iFFT transformation or a too low resolution. Due to this pronounced different, it has been decided to only compare physiological conditions for each logging method separately.
Figure 6.1: A delay in pulse can be observed. The plot shows the velocity profiles (averaged over 7 scans) of 6 gridpoints (roughly) positioned on a vertical line. A delay of $t_2 - t_1 = 25\text{ms}$ can be observed between the upper and lower gridpoints. Assuming a distance of approximately 10 to 15 cm, yields a wave velocity ranging between $4$ to $6 \text{ m/s}$, a similar order as Aortic Pulse Wave Velocities reported in literature.

6.2 Influence spatial distribution

As previously observed in section 5.4.2, skin motion appears to be dependent on gridpoint location. Straight above the Abdominal Aorta, the (averaged) profile shows a relative simple ‘cardiac sine’, further away the signal increases in complexity (figure 6.3 and also observed in figures 6.4, 6.5). However, closer inspection also reveals several other artifacts which can be attributed to the location of the measurement position on the abdominal surface.

Inspection of the samples suggests that the ‘cardiac sine’ tends show a delay when comparing velocity profiles of gridpoints positioned along the y-direction. This observation is consistent the presence of a pressure pulse along the cardiovascular system. Due to the finite stiffness or the arteries, the changes in pressure travel with a finite Pulse Wave Velocity (PWV) through the endovascular system. Depending on age, reported values in the aorta (between carotid and femoral artery) range between 5 to $12 \text{ m/s}$ for normal subjects with low prevalence of hypercholesterolemia [Avolio et al. 1983]. Detailed analysis of the delay of the cardiac sine observed in gridpoints positioned in the vertical line above the abdominal aorta, shows a pulse delay of 25 ms, which (assuming a distance of approximately 10 to 15 cm between upper and lower gridpoint), yields a PWV ranging between 4 to $6 \text{ m/s}$ (figure 6.4). This value is in a similar order as the PWV values reported in literature, and strengthens the assumption that the observed cardiac sine is directly related to the dilation of the abdominal aorta.

Furthermore, the velocity profile appears to be affected by underlying tissue. A number of scans have been performed on a higher resolution grid, defining a larger number of gridpoints on the
abdomen and also placing several tents of gridpoints on the lower part of the thoracic cage. Inspection of the velocity profiles positioned above the thorax reveals that these locations experience a completely different vibration. Although it is still possible to distinguish a pulse, its amplitude is considerably smaller (at least twice as small), shows less complexity and also appears on a later time instant as observed when measuring above soft tissue (figure 6.2). This could indicate a different transfer mechanism between heart activity and skin vibration or maybe even an other input source (for instance, it is possible to palpate the so-called apical pulse on the thorax, which is caused by direct contact between heart and chest wall. This mechanism could be the responsible input source for this particular vibration pattern).
Figure 6.2: A detailed look on the velocity profiles acquired during scanning of both the abdomen and part of the lower thorax. Figure 6.2(a) shows the velocity profiles of the gridpoints associated with line L1, which lies on the thoracic cage, acquired during 3 separate scans. Figure 6.2(b) shows the velocity profiles of gridpoints lying in the center of the abdomen. Figure 6.2(d) and 6.2(e) show the profiles of the gridpoints lying on the right and left of the midpoints respectively. Figure 6.2(c) shows the gridpoint locations of the velocity signals plotted in subfigures 6.2(a), 6.2(b), 6.2(d) and 6.2(e). It can be observed that on the location lying above the thoracic cage, the vibration amplitude is significantly smaller, is less complex and the cardiac pulse appears at a later instant, after 200 - 400 ms, while the pulse of the midpoints appears almost immediately.
6.3 Stomach Content

In order to evaluate the influence of stomach content, the subject was measured (on two occasions) with a sober stomach and after eating breakfast. During the first session (tables 6.1 & 6.2) the results where logged as FFT spectra, during the second session (table 6.3) the results where logged in a velocity - time representation. Because each session used a different array of gridpoints and data log methods, both are compared in separate plots (figures 6.3 and 6.4). It appears that the velocity profiles logged as an FFT does give a somewhat different result, appears to have a different scaling than the direct velocity-time representation and also shows a different location of the cardiac sine. The origin of this difference is not known.

On overall the shape of the averaged velocity profiles remains fairly similar in shape, so it does not appear that stomach content changes the global cardiac induced vibration at a very distinct matter. However, it does appear that amplitudes of the cardiac sine are slightly reduced after eating dinner. This effect is most prominently seen in the velocity profiles of 6.4 at the upper midst of the abdomen, which is indeed the area where the stomach should be present. However, there are not sufficient scans available to conclude if this conclusion can be drawn.

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<th>5 &amp; 12</th>
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<td>100</td>
<td>100</td>
</tr>
<tr>
<td>Samp. Time(ms)</td>
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<td>1000</td>
<td>1000</td>
</tr>
<tr>
<td>Bandwidth</td>
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<tr>
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<td>BP 0.5 - 50 Hz</td>
<td>BP 0.5 - 50 Hz</td>
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<td>DD-200</td>
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<td>Velocity (mm/s/V) LP</td>
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<td>640µm/V</td>
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<td>4 * complex</td>
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Table 6.1: Settings used during the measurements in session 4. LP = Low Pass filter, BP = Band Pass filter. VD = Velocity Decoder, DD = Displacement Decoder.

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<tr>
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<td>no aver.</td>
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</table>

Table 6.2: Settings used during the measurements in session 4. LP = Low Pass filter, BP = Band Pass filter. VD = Velocity Decoder, DD = Displacement Decoder.
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**Table 6.3:** Settings used during the acquisition of dataset S7: From test 5 onwards the QRS block wave was also recorded on REF1, making it possible to resize the sample window in order to synchronize with the cardiac cycle.
Figure 6.3: Individual averaged velocity profiles for each gridpoint. Blue is sober, Red is full stomach. The cross corresponds with the location of the umbilicus. A photo of the gridpoint on the abd. surface can be found in figure 6.2(c).
Figure 6.4: Individual averaged velocity profiles for each gridpoint. Blue is sober, Red is full stomach. The cross corresponds with the location of the umbilicus. A photo of the gridpoint on the abd. surface can be found in figure 6.5.
Figure 6.5: Location of the gridpoints of the grids used in figure 6.4.
6.4 Respiration

Literature (section 4.2.1) describes the respiration to be source in the abdomen surface motion and revealed that its influence should be restricted a very low frequency band (0.1 - 0.3 Hz). The performed measurements indeed show low frequent signal on the measurement, which is believed to be the result of respiration.

Due to the low frequency of the respiration motion, it is associated with relatively small velocities. Its presence dominates the displacement (yielding visible displacements in the order of cm’s), but is only partially visible in the velocity profile of the abdomen surface as an low frequent signal as figure 6.6 illustrates. Integration of the velocity signal does reveal that a relative small offset in the velocity signal yields large displacements (figure 6.7), resembling the displacement patterns previously measured when logging displacement of the abdomen surface (figure 4.10; please note that the displacement shown in the latter figure where logged after four times complex averaging of a FFT spectra and, due the effect of averaging on the unsynchronized respiration cycle, is somewhat distorted. The results shown in figure 6.7 are integrated out of single velocity acquisitions.)

More disrupting is the tendency of the respiration to cause the laser beam to move over the reflector surface, or even worse, to cause a ‘loss of focus’ because the laser beam does not hit the reflector anymore. The first event introduces ‘speckle noise’, which can be partially resolved by using a Speckle tracking filter (an option within the LDV software, which bypasses the intermittences caused by speckles by means of a Phase-locked Loop system [Polytec Software Manual]). The second event is more disruptive, because the beam will be back-scattered by the skin dermis, hair follicles or hairs, generating longer duration, high amplitude noise which cannot be eliminated by means of tracking filters. More details regarding the effect of noise on signal quality can be found in appendix B.5.2 and B.5.3.

![Velocity profile of successive gridpoints](image)

**Figure 6.6:** Velocity profile of the successive gridpoints 9 till 12 showing a low frequency trend occurring in the velocity profile. The frequency of this trend appears to be about 0.15 - 0.25 Hz, and is most probably caused by respiration. The data is retrieved from dataset S6V6. The It should be noted that each sample window is triggered by the occurrence of a QRS wave; the time period between two sample windows can vary considerably, causing discontinuities.
Figure 6.7: The influence of respiration is restricted to low frequencies. As such, its influence is mostly visible in the displacement and less visible in the velocity profile. Above figure illustrates this effect: integration of velocity profiles (left) yield displacement profiles (right) which hardly show any cardiac cycle related motion. Note the resemblance between the integrated profiles and previously measured displacement signals (figure 4.10).
6.5 Muscle activity

In order to evaluate the influence of muscle activity, the subject was asked to tighten his abdominal muscles during the duration of one single scan. As to be expected, tightening the muscles causes a large displacement of the abdomen surface, causing all of the reflectors to move out to the focus position of the laser. The obtained velocity measurements show a significant degree of noise. However, application of a digital moving average filter (with a moving average window of 32 samples or 63 ms) seems to clear most of the noise. It can be observed that the filtered signal (figure 6.8) does not show the appearance of the earlier observed ‘cardiac sine’, and shows a fair amount of vibration, appearing to have a frequency bandwidth around 10 Hz.

![Velocity-time graphs](image)

**Figure 6.8:** Tightening of the abdominal muscles results in a large displacement of the abdominal wall, causing the reflectors to move out the the focus of the laser beam. The loss in reflected signal power reduces the signal to noise ratio, increasing the impact of noise. Introducing a digital moving average filter (with windowlength of 32 samples) during post-processing appears to remove most of this noise. The filtered signals show increased vibrations with a bandwidth of approximately 10 Hz.
6.6 Pulsation of abdominal aorta

The cooperation with the LUMC resulted in the unique possibility to perform measurements on patients with an real Abdominal Aortic Aneurysm (AAA). An AAA is a localized dilation of the abdominal aorta, mostly occurring below the kidneys and above or partially among the iliac arteries in the pelvis [Wikipedia Abdominal Aortic Aneurysm]. Although this condition is often asymptomatic, the expansion and weakening of the lumen wall usually results in increased pulsation of the aneurysm, and, at a certain degree, can be detected using palpation [Fink et al. 2000]. It is assuming that such a strong physiological event would also influence the internal dynamics of the abdomen, and as such, should influence the motion of the skin surface. With the help of Dr. Jan-Willem Hinnen, measurements where performed at the LUMC in order to measure skin motion on real life patients who have been diagnosed with AAA and where scheduled for Endovascular Repair (EVAR).

Results are shown for two different subjects, the first patient was scheduled for the placement of a mono-iliac stent-graft, the second patient was scheduled for a bi-iliac stent-graft. Abdominal vibration was measured prior to the EVAR operation and several days after the procedure. The motion of the abdomen was recorded by measuring the velocity of retro-reflective stickers. A Finometer was included to measure the brachial blood-pressure waveform of the subjects (figure 4.7(a)) and its results were stored by the LDV.

6.6.1 Velocity profile

Figures 6.10 and 6.11 show the velocity profiles of the patients prior to EVAR. Comparison with the velocity profiles previously shown for a healthy young subject (figure 6.4) shows a remarkable increase of pulsation. While the cardiac pulse of the healthy subject was only clearly visible above the abdominal aorta, the pulsation of the AAA patients appears be present at a larger area. The velocity also shows a clear increase in amplitude.

![Figure 6.9](image)

**Figure 6.9:** Photo's of the abdomen showing the position of the gridpoints on the abdominal surface. Photo 6.9(a) shows the grid used in 6.10. Photo 6.9(b) shows the grid used in 6.11.
Figure 6.10: Plot of AAA patient 1 prior to EVAR. The thick blue profile shows the average velocity profiler averaged over all of the scans. The red shows the t-student calculated 90 % confidence interval around the average. The green line shows the standard deviation around the mean values. The blue cross corresponds with the location of the umbellicus. The photo shown in figure 6.9(a) shows the position of the gridpoints on the abdominal surface.
Figure 6.11: Plot of AAA patient 2 prior to EVAR. The thick blue profile shows the average velocity profiler averaged over all of the scans. Again the red line indicates the 90% confidence interval, the green line shows the standard deviation. The blue cross corresponds with the location of the umbilicus. The photo shown in figure 6.9(b) shows the position of the gridpoints on the abdominal surface.
6.6.2 Frequency Response Functions

Similar to the procedure described in section 5.6.2, Frequency Response Functions can be defined between the input (brachial blood pressure waveform) and output signals (velocity). Again the derivative of the blood pressure \( \frac{dP}{dt} \) is used as input in order to eliminate the DC offset present within the pressure waveform.

Figure 6.12 shows both the frequency responses and the coherence functions of patient 1. Both functions have been plotted for each of the gridpoints after excluding several particular noisy signals (corresponding to scanpoints where the reflector moved out of the laser beam). The FRF for each gridpoints is obtained after averaging Autopower spectra and Crosspower spectra of each acquired datasets (in this case datasets V1 - V4, each comprised of a single completed abdomen scan), as defined in equation D.29 [appendix D.4.2]. It can be observed that the FRFs appear quite similar. However, the coherence function is fluctuating a lot, indicating that the assumed input signal is not the only source of the skin motion. There are several factors which could lead to this phenomenon.

Both the velocity as the blood pressure signals have been recorded on relative small sample lengths, influencing the ability of the FFT to capture all of the characteristics of the signal, even more when considering additional losses caused by the application of the window functions. It remains a possibility that the inputs for the FRF does not sufficiently contain the characteristics which excite skin motion while the results are present within the output signals of the FRF, or vice versa.

Figure 6.13 shows the FRF and coherence functions of patient 2, plotted for each scanpoint. Comparison with the FRF shown in figure 6.12(a) shows that the FRFs appear to have a similar shape, but this time slightly clearer than previously observed. The coherence function is quite different, showing quite high coherence (between 0.8 till 0.95) between 5 and 9 Hz. It is interesting to note that the FRF appears to show resonance peaks at 5 and 9 Hz, but this is also accompanied with a sudden drop in coherence. Anti-resonance usually causes a drop in coherence due to lower signal to noise ratio at these ratios. Coherence functions that show dips at resonance, generally indicate a leakage problem [Heylen et al. 1998].
(a) Frequency Response Functions (for each scanpoint) from the pre-EVAR measurements performed on patient 1.

(b) Coherence functions (for each scanpoint) from the pre-EVAR measurements performed on patient 1.

Figure 6.12: FRF and Coherence functions of the pre-EVAR measurements performed on patient 1. The FRF’s and Coherence functions are calculated using the average APS’s and CFS’s over datasets [V1, V2, V3]. The plot excludes the FRF’s of scanpoints 3 and 25, which appeared to experience noise during acquisition. The derivative of blood pressure $\frac{dP}{dt}(t)$ was used as input function, the velocity signal $v(t)$ was used as output. A Hann window function was used on $\frac{dP}{dt}(t)$, a (10% - 10%) Tukey window was used on $v(t)$. 
Figure 6.13: FRF and Coherence functions of the pre-EVAR measurements performed on patient 3. The FRF’s and Coherence functions are calculated using the average APS’s and CFS’s over datasets [V1 - V7]. The plot excludes the FRF’s of scanpoints 4, 8, 27, 33, which appeared to experience noise during acquisition. The derivative of blood pressure \( \frac{dP}{dt}(t) \) was used as input function, the velocity signal \( v(t) \) was used as output. A Hann window function was used on \( \frac{dP}{dt}(t) \), a (10% - 10%) Tukey window was used on \( v(t) \).
6.7 Conclusions

Tests are performed to evaluate the influence of several easy controllable physical parameters: e.g. the influence of respiration, the difference between sober stomach and full stomach, influence of tensioning abdomen muscles and, quite unique, the influence of increased Abdominal Aorta pulsation the vibration experience on the abdomen surface.

As previously observed in section 5.4.2, the skin above the abdominal aorta shows a clear distinct cardiac pulse. Closer inspection reveals that a small delay of the cardiac pulse develops between gridpoints positioned further away from the hearth. Calculation shows that this delays appears to be in the range of 4 to 6 fracms, a similar order as literature reports for Pulse Wave Velocities for healthy aorta’s. The biomechanical structure lying under the gridpoints significantly influences the vibration of the skin surface. Gridpoints positioned on the thoracic cage show a delayed ‘cardiac sine’ with smaller amplitude. It would be reasonable to assume that vibrations caused by direct contact between the heart and the thorax are transmitted directly through the thoracic cage. However, due to the stiffer nature of this path, this would also imply faster transmission, which counteracts the observation of a delay. Because research for an explanation for this discrepancy lies beyond the scope of this research, it is just concluded that the motion of gridpoints positioned above the stiffer thorax are most likely not representative of internal abdominal vibrations, and scans should be limited to the soft parts of the abdomen.

Comparing the results between sober and full stomach shows fairly similar patterns, although the cardiac pulsation at the location of the stomach does appear to show a slight reduction of amplitude. However, the number of measurements performed are not sufficient to support or disproof this hypothesis.

Although not immediately perceived in the velocity measurements obtained during the initial tests, it was shown that respiration significantly alters skin motion. This problem is attributed to the fact that the initial measurements were logged as an FFT spectrum. This method, which does not log a zero DC component and due to its application of complex averaging prior to saving the results, tends to underestimate the influence of respiration on the velocity profiles. However, by calculating an average velocity profile over multiple scans, it is indeed possible to average its influence out of the signal. Respiration does increases the likelihood of the reflectors to move out of the beam focus, which can be solved by using sufficiently large reflectors.

Tensioning of the abdominal muscles causes a large deflection of the abdominal wall, usually causing the reflective stickers to move out of the focus of the laser beam. Filtering does give a reasonable signal, which does not show any presence of the characteristic ‘cardiac’ sine, but instead a increase of random vibrations.

Measurements on AAA patients revealed a significant increase of vibration amplitude, also suggesting that pulsation of the abdominal aorta is indeed a major excitation source. While the standard subject showed an vibration amplitude in the order of 5E-3 m/s above the abdominal aorta, AAA patients appear to have vibration amplitudes that are at least twice as large, ranging up to 25 E-3 m/s. The pulsation also appears to cover a larger area and shows different a different shape than previously observed.
Chapter 7

Conclusions

The research showed that it is indeed possible to use a Laser Doppler Vibrometer to detect vibration of the abdominal wall. It is however necessary to place retro reflective stickers on the abdominal surface in order to obtain sufficient signal quality. The light frequency used in the LDV laser has got a tendency to penetrate quite deep into the skin before being backscattered. During this process, hair follicles and other irregularities contained within the skin can cause noise, which due to the low signal power returned to the detector, can significantly distort the velocity signal. Although it is possible to measure both displacement as velocity, it is highly recommended to restrict to velocity measurements because this type of measurements are less sensitive for the occurrence of dropout.

Although the system is not able to perform a parallel scan, triggering on the QRS complex of the heart results in fairly repeatable patterns, in which it seems possible to distinguish pulsation of the abdominal aorta. Pulsation is most clear in the gridpoints lying above the abdominal aorta, showing a clear ‘cardiac sine’, whereby it even appears to be possible to distinguish the Pulse Wave Velocity (PWV) of the abdominal aorta.

Cardiac vibration and cardiac induced sounds are reported to be mostly present in a bandwidth of 0.5 - 50 Hz, although the results show that abdomen vibration is most dominantly present between 0 - 25 Hz. Respiration will cause organ motion of the abdominal organs, up to several centimeter along the superior - inferior axis and 1 - 2 mm in the anterior - posterior direction. However, motion of the diaphragm (caused by respiration) will cause the abdominal cavity to bulge out, such that skin motion due to respiration will be primarily in the anterior - posterior direction. Respiration is present as a low frequent (0.15 - 0.2 Hz) distortion within the abdominal surface velocity, but can be reduced by averaging the results of multiple scans. Excitation sources like respiration sounds and bowel sounds are expected to lie above 100 Hz and have not been observed in the measurements.

Evaluation of the measurement results revealed that skin velocity patterns show quite a lot of spatial variation and it is believed that it is necessary to use analysis techniques that are able to distinguish spatial position during evaluation of skin velocity patterns. Averaging the signals over abdomen regions is believed to oversimplify the interpretation and it is believed this results in the loss of the subtle features within the surface vibration patterns that could be related to local mechanical perturbations.

Analysis techniques like frequency response functions, operational deflection shapes and 2D spatial Fourier transforms are useful techniques to assess dynamic behavior of abdominal surface. Frequency response and coherence functions appear to show a dynamic response of the abdominal surface under influence of the blood pressure waveform, suggesting that the motion of the abdominal surface is partially caused by dynamic processes.
Chapter 8

Future work & Recommendations

8.1 Future work

At the moment no follow up research has yet been planned, nor has a budget been allocated to fund additional research. However, the Delft University of Technology is currently in the process of applying for a patent in order to protect the methodology and accompanying measurement setup. If everything goes well, it is hoped that additional research partners can be found in order to continue research and further improve the methodology and measurement setup.

The author and his coordinators plan to publish an article on relative short term, which will discuss the methodology and results in more detail.

8.2 Recommendations

For sake of clarity, the recommendations are split up into recommendations regarding possible improvements of the measurement setup and suggestions to improve the vibration analysis. The first part primarily focuses on technical details that would improve the setup, the latter consist basically out of suggestions for further research, which can be performed with the current setup.

8.2.1 Suggested improvements of the measurement setup

The current measurements are distorted by respiration, which is removed by averaging over multiple scans. It is highly recommended to implement a faster and more direct method of removal, because direct identification of respiration induced motion would make it possible to reduce the amount of scans necessary in order to obtain usable results. It is suggested to implement of some sort of feedback or tracking system: by distinguishing the motion caused by respiration, it should be possible to subtract the motion induced by respiration from the abdomen surface motion, yielding a direct representation of cardiac cycle induced motion.

The current LDV requires the placement of reflective sticker on the abdomen surface, requiring some preparation before it is possible to start scans. Follow up research should include testing of (diffuse) reflecting materials which can be applied more easily and faster. However, care should be taken that those materials do not pose a health risk when applied on human tissue. Possible solutions could be the use of CADCAM sprays such as used in laser imaging of denture features. Other possible methods could be wearing tight reflective clothing or the implementation of alternative laser frequencies which show improved signal reflection on skin tissue.
Scans are currently triggered by the presence of the QRS complex, monitored by an ECG monitor. However, any cardiac related event which sufficient accuracy could be used to trigger the scan, providing an opportunity to simplify setup. Instead of monitoring with a three lead system, it may be possible to simplify the system, maybe by using a similar systems as used for heart rate measurement in fitness equipment. It could also be an option to register heartbeats by using a separate laser to measure thorax vibration (for instance by monitoring chest vibration at the apex of the heart).

The current measurement setup uses an expensive industrial grade Laser Doppler Vibrometer, designed to measure a wide range of frequencies and resolutions. Significant cost reduction should be possible by downsizing of the equipment towards the requirements necessary for abdomen scanning.

It is highly recommended to try to quantify motion by measuring multiple gridpoints in parallel instead of relying on a sequentially obtained approximation. There are already commercial systems available which are able to measure multiple points in parallel. Another option could be the use of photogrammetry as an alternative for laser, as this would enable the instantaneous tracking of multiple gridpoints. Implementation of such an improvement on the system would reduce scanning time and would make it possible to measure multiple points in a similar phase of respiration.

8.2.2 Suggested improvements on the analysis of Abdomen vibrations

This research project was started with the intention to provide a proof of concept and measurements where restricted to a very limited set of volunteers. Although these measurements gave some insight of intra-patient variability, no effort has yet been taken to test inter-patient variability. The next step would be to observe intra-patient variability on other subjects and to assess the amount of inter-patient variability between patient in similar physiological condition.

A finometer was implemented to obtain an approximation of a vibration excitation source input by monitoring the brachial blood pressure waveform. When defining a frequency response function between finometer input and abdomen vibration output, this also includes the transfer function between the brachial blood pressure waveform and the pulsation of the abdominal aorta. Literature mentions that aging and cardiovascular diseases can alter the relation between the central and brachial pressure waveforms. Further research is necessary to assess this effect on the definition of a frequency response function and if it is possible to identify and isolate this phenomenon.

Analysis of the abdomen motion patterns showed the emergence of repeatable vibration patterns which do not show resemblance to the simple ‘cardiac sine’-like pulsation which appears above the abdominal aorta. They do however appear to be synchronized with the cardiac cycle, suggested that they could originate from the same excitation source. More research is required to determine the origin of this complexity.

Measurements on AAA patients showed a significant increase of vibration amplitude. Additional research is required to determine if this increase originates from an increase in pulse pressure, a reduction of abdominal aorta wall stiffness or if it caused by patient-specific anatomy.


Appendices
Appendix A

LDV Operation

A.1 Introduction

A Laser Doppler Vibrometer (LDV) is able to measure vibrations of a surface without the need of contact between the sensor and the surface. The sensor operates on the principle of laser Doppler interferometry; a laser beam with a known frequency is directed at the surface of interest, reflects at the surface and is recorded by the sensor. Movement of the surface will result in a Doppler shift of the reflected beam frequency, dependent on vibration amplitude and frequency. By determining the frequency shift, it is possible to extract the target velocity component along the direction of the laser beam.

A.2 Doppler interferometry

A vibrometer is generally a two beam laser interferometer that measures the frequency (or phase) difference between an internal reference beam and a test beam. The test beam is directed to the target, and scattered light from the target is collected and interfered with the reference beam on a photo detector, typically a photo diode. Most commercial vibrometers work in a heterodyne (using 2 dissimilar light frequencies) regime by adding a known frequency shift (typically 30-40 MHz) to one of the beams. This frequency shift is usually generated by a Bragg cell, or acousto-optic modulator.

A schematic of a typical laser vibrometer is shown in figure A.1. The beam from the laser, which has a frequency $f_o$, is divided into a reference beam and a test beam by a beam splitter. The test beam passes through a Bragg cell, which adds a frequency shift $f_b$. This frequency shifted beam then is directed to the target. The motion of the target causes a Doppler shift to the beam given by equation A.1. It should be noted that the Doppler shift is caused by the velocity component parallel to the beam direction, hence the $\cos (\alpha)$ within relation A.1. Wikipedia Laser Doppler vibrometer, Polytec CD-ROM.

\[
 f_d = 2 * v(t) * \cos (\alpha) / \lambda
 \]  

(A.1)

Where:
$v(t)$ is the velocity of the target as a function of time;
$\alpha$ is the angle between the laser beam and the velocity vector;
Figure A.1: Basic Components of a laser Doppler Vibrometer. A beam splitter generates a reference beam and a test beam. The test beam passes through a Bragg cell and is directed at the to be measured surface, which will cause a Doppler shift on the signal. The returned test beam will be added to the reference beam, causing an interference pattern. The change in intensity of the interference pattern (at its so called beat frequency) is sufficiently slow to be detected by the photo detector [Wikipedia Laser Doppler vibrometer].

\[ \lambda \] is the wavelength of the light.

Light scatters from the target in all directions, but a portion of the light is collected by the LDV and reflected by the beam splitter to the photo detector. The frequency of this light \((f_o + f_b + f_d)\) contains the frequency components of the laser, Bragg cell and the Doppler shift. The collected light is combined with the reference beam at the photo-detector and will generate an interference pattern. While the initial frequency of the laser is too high to be detected by a photo detector (red laser light will have a frequency in the order of \(10^{14}\) Hz), the beat frequency \(f_b + f_d\) of the interference pattern will be much lower, in the order of tens of MHz, and can be detected.

The output of the photo detector is a standard frequency modulated (FM) signal (around 40 MHz), with the Bragg cell frequency as the carrier frequency, and the Doppler shift as the modulation frequency. This composite signal is demodulated using I/Q signal generation into two orthogonal components, as described in equation (A.2)

\[
\begin{align*}
  f_b + f_d &= A(t) \sin [2\pi ft + \phi(t)] = \\
  I(t) \cdot \sin (2\pi f) + Q(t) \cdot \sin (2\pi f + \frac{\pi}{2}) \\
  I(t) &\cong A(t) \cos [\phi(t)] \\
  Q(t) &\cong A(t) \sin [\phi(t)]
\end{align*}
\]

The resulting I and Q signals can be represented as a rotating vector, shown in figure [A.2]. The angle of the rotating vector corresponds with the phase of the signal (corresponding with the displacement of the surface). The rate of angle chase (e.g. the angular velocity of the rotating vector) corresponds with the change in phase of the signal, proportional to the surface velocity. The length of the vector indicates the power of the light hitting the detector, and is independent of vibration velocity or displacements (However, signal power is still important for noise reduction, as will be explained in appendix [B]).
Figure A.2: Representation of the I/Q demodulated signal as a rotating vector. The length of the vector corresponds with signal power and is independent of vibration characteristics. The change in phase angle $\varphi$ corresponds with displacement, the rate of change in phase angle $d\varphi/dt$ corresponds with velocity.

A.3 Transfer function LDV frame

In order to check if the structure of the measurement could introduce noise within the measurements, the dynamics of the frame have been analyzed by means of an excitation experiment. An acceleration sensor was positioned on the location of the LDV head and the frame was excited in X, Y and Z direction by means of a impulse hammer. For each direction a transfer function is defined and plotted in figure A.3 showing both amplitude and phase of the obtained transfer functions.

Most resonance frequencies appear to lie above 100 Hz, which should not interfere with the investigated frequency band, but there also appears to be resonance present at 20 Hz. Unfortunately the direction of the acceleration sensor was not stored, so it is impossible to distinguish the direction of the transfer functions.
Figure A.3: Transfer functions of the LDV frame in X, Y and Z directions. Most resonance peaks lie above 100 Hz, but there also appears to be resonance present at 20 Hz. This resonance does coincide with the bandwidth of interest and could indeed interfere with the measurements.
Light transmission

B.1 Introduction

An important aspect of this research is the amount of noise attained during the Laser Doppler Interferometry use. In order to evaluate this feature, this appendix will treat the optics of the measurement system in more detail.

B.2 Basic light theory

B.2.1 Scattering, reflection and refraction

Scattering is a general physical process where radiation (such as light), is forced to deviate from a straight trajectory by one or more localized non-uniformities in the medium through which they pass. In conventional use, this also includes deviation of reflected radiation from the angle predicted by the law of reflection. Reflections that undergo scattering are often called diffuse reflections and unscattered reflections are called specular (mirror-like) reflections.

Specular reflection behave in a simple, predictable way. Diffuse reflections occur at matte surfaces and can only be described statistically, with the exact distribution of the reflected light being dependent of the microscopic structure of the material. Refraction is the change of light direction due to a change in refraction index, for example incident light on an air-water boundary under an angle. Lenses use this principle in order to focus light.

When radiation is only scattered by one localized scattering center, this is called single scattering, if scattering centers are grouped together, such that the light may scatter many times, is called multiple scattering. Single scattering can be treated as an random phenomenon, multiple scattering is usually more deterministic due to the averaging effect of the large number of scattering events.

Light scattering can be divided into three domains based on a dimensionless size parameter, $\alpha$ which is defined as:

$$\alpha = \frac{\pi D_p}{\lambda}$$  \hspace{1cm} (B.1)

where $\pi D_p$ is the circumference of a particle and $\lambda$ is the wavelength of incident radiation. Based on the value of $\alpha$, these domains are:

$\alpha \ll 1$: Rayleigh scattering (small particle compared to wavelength of light)
\( \alpha = \frac{1}{M} \) : Mie scattering (particle about the same size as wavelength of light)
\( \alpha \gg 1 \) : Geometric scattering (particle much larger than wavelength of light)

Rayleigh scattering is a process in which electromagnetic radiation (including light) is scattered by a small spherical volume of different refractive index, such as a particle, bubble, droplet, or a density fluctuation. The Rayleigh’s model applies on small particles: the diameter of the sphere is much smaller than the wavelength of the scattered wave; the upper limit normally taken to be about 1/10 the wavelength. At this microscopic scale, the exact geometry of the scattering center is not very important and can usually be simplified as a sphere of similar volume. Radiation passing through a pure gas is scattered due to the microscopic density fluctuations caused by the motion of the gas molecules. These molecules are small enough for Rayleigh’s model to apply. The magnitude of scattering is mainly influenced by the ratio of particle diameter to the wavelength of the radiation, but is also a function of polarization, angle, and coherence.

In the Mie regime, the shape of the scattering center becomes much more significant and the theory only applies well to spheres and, with some modification, spheroids and ellipsoids. Closed-form solutions for scattering by certain other simple shapes exist, but no general closed-form solution is known for arbitrary shapes.

At larger scales, geometric scattering can be described by geometric optics (ray optics) describing light propagation as rays, propagating in a straight line. Rays can experience reflection or refraction, and can be described by the Law of Reflection and Snell’s law [Wikipedia Optics].

**B.3 Laser to skin**

The laser beam will experience some scattering in its path from laser source to the surface due to Rayleigh scattering caused by the molecules of the gas, but at the path lengths under consideration, this effect can be neglected. Another more significant distortion can be caused by water droplets in the air, causing Mie scattering, but this effect can be contained by using proper climate control within the research area. Excessive dust can also have some effect on the amount of light transmitted to the surface.

**B.4 Skin Optics**

Light hits the skin surface and is partially redirected towards the detector. However, it should be realized that skin is a biological tissue, and as a result, shows quite complex optical properties. Skin is composed out of several distinct layers, all having different optical properties. Figure B.1 shows a cross-section of the skin, identifying the following layers:

**Stratum Corneum** The outermost layer of the skin, composed of flat, keratin filled plate-like envelopes, consisting out of dead cell originating from the epidermis. This tissue shows no light reflection, transmission through the layer is approximately uniform for visible light. Only a small fraction (4 % to 7 %) of the incident light is reflected due to the change in refractive index between air \( (n_D = 1.0) \) and stratum corneum \( (n_D \approx 1.55) \) [Anderson and Parrish [1981]]. Due to the non-smooth and planar surface the regular reflectance is not specular (mirror-like) and light passing through this layer will be made more diffuse due to refraction caused by the rough interface. An increased reflection is experienced in psoriatic plaques, where excessive amounts of skin flakes results in additional optical interfaces, resulting in a white, scaly appearance.

**Epidermis** The epidermis, combined with the stratum corneum, forms the most outer skin layer. Although containing several specialized skin cells, like melanocytes, it does not contain any
skin organs or blood transportation. Combined thickness is somewhere between 10 - 150 µm, although it can be as thick as 1.5 mm on the palms and soles. The thickness of the stratum corneum and epidermis is still small enough such that its contribution to remittance is minimal over the entire visible and near infra-red spectral regions. The ratio of diffuse / direct transmission is almost independent of wavelength, suggesting that particle scattering is not the main process for epidermal transmittance. It is mostly transmitting or absorbing, it is most absorbing for small wavelengths (UV light): besides the remittance due to regular reflectance at the air-skin interface, only 5% of the collimated incident radiation in the 300-3000 nm region is remitted by scattering within Caucasian epidermis. Epidermis is thin enough such that its scattering properties can be neglected compared with the dermis.

**Dermis** Scattering is important in the dermis. Longer wavelengths penetrate deeper into the skin than shorter wavelengths (on UV-visible-near infrared radiation). Light of 600 - 1300 nm penetrates quite deep into skin and human tissue in general (the so called ‘optical window’). Total absorption coefficient within the dermis is quite dependent of hemoglobin absorption. Scattering consists out of Mie scattering by the large cylindrical dermal collagen fibers and the Rayleigh limit scattering by the small-scale structures associated with collagen fibers and other cellular structures. Scattering of visible to near-infrared light is affected by Mie and Rayleigh scattering, both mostly the result of (small-scale) collagen fibers.

As mentioned above, skin remits visible light mainly due to scattering within the skin layers, causing a diffuse, but quite light appearance. The change in index of refraction at the air-skin interface causes (diffuse) light reflection, but only accounts for 4% to 7% of overall re-emitted light. The epidermis is mostly transmitting and absorbing, where melanin exhibits a monotonically decreasing absorption coefficient with increasing wavelength and primarily absorbs UV radiation. Light above 600 nm, like the 630 nm used in the commercial Polytec laser [Cite + check brochure!], is mostly scattered instead of absorbed. It also penetrates deeper into the skin (it can be stated that longer wavelengths result in deeper penetration), into the dermis. [Desjardins et al. 2007, Gemert et al. 1989]

![Figure B.1: Layout of the skin. (a) shows the layout of human skin, (b) shows the skin model generally used in skin optics.](image)

**B.5 Skin to detector**

After being reflected / backscattered at the skin surface, part of the reflected light returns optical sensor of the Laser Doppler Vibrometer. Similarly to the beam traveling from laser source to the
surface, the light will encounter scattering due to Rayleigh and Mie scattering caused by the air medium. More interesting is to considering some of the effects accompanied by the geometrics of the scanning surface and the influence of returned signal power on signal quality.

### B.5.1 Scan Geometrics

Since the LDV only detects vibration velocity in the direction parallel with the laser beam, the effect of curvature of the and location of the LDV scanning head influences the measurements to some extend. This effect is illustrated in figure B.2. An estimation based on the geometry of the setup during session 1 ($L_1 = 710\text{mm}, L_2 = 730\text{mm}, L_3 = 750\text{mm}, W_2 = 160\text{mm}$) results in underestimation of $V_a = 0.97V_{r,a}$ and $V_b = 0.95V_{r,b}$ (assuming a ‘flat’ abdomen surface). Positioning the tripod close to the abdomen surface and maximizing the distance between scanning head and the to be scanned surface, should keep error below 10%.

\[
V_{r,a} \approx V_a \cdot \frac{L_1}{L_2} \\
V_{r,b} \approx V_b \cdot \frac{L_1}{L_3}
\]

The curvature of the abdomen surface will cause an additional error: using a tripod will cause additional error on the far side of the abdomen when the curvature increases (for instance when the subject is obese).

![Figure B.2: The position of the LDV scanning head influences the results of the measurements, depending on the precise geometry. The error due to geometry is the most severe when applying a tripod, because this setup will place the scanning head partially besides the abdomen surface. In the situations encountered during acquisitions, the dimensions of $L_1, L_2, L_3, W_1$ where such that the error between the detected velocity $V_a$ (parallel to the laserbeam) and the real velocity $V_{r,a}$ (perpendicular to the surface) is at most 10 %.](image)

### B.5.2 Influence of return signal power on signal quality

As explained in the previous appendix summarizing the operation principles behind Laser Doppler Vibrometry (Appendix A), the optical signal is demodulated into a $I/Q$ signal and can be represented as a rotating vector, where the angel represents the displacement of the surface and the angular velocity of the rotating vector corresponds with the velocity of the surface. The length of the vector represents the signal strength, and does not contain any information regarding vibration...
velocity or displacement; in optimum conditions it can be stated that the measurement principle is insensitive for changes in reflected signal.

However, in practice, dropouts occur during demodulation, influencing the complete signal as shown in figure B.3. The complete signal can be seen as the sum of the ‘perfect signal’ and noise. Assuming that noise does not have any preferred phase, the resulting vector is randomly influenced by the noise component. If sufficient signal power is returned, the influence of the noise component is limited. However, at moderately low signals, noise becomes important and can cause considerable variation in phase, yielding variation in the angular velocity, causing a noise velocity signal. At very low signal, the noise can even cancel out the signal. The random nature of noise phase can change very rapidly, causing large peaks in the velocity, resulting in the appearance of dropouts within the signal [Polytec CD-ROM].

![Figure B.3: Influence of signal power on signal quality represented as a rotating vector.](image)

The perfect signal (green) corresponds with a displacement $\varphi$ and angular velocity $d\varphi/dt$. When measuring a (realistic) surface vibration, the angle $\varphi$ should change smoothly. Depending on signal strength, the random phase of the noise (orange) will affect signal quality. With good signal power, the addition of noise (orange) hardly effect the resultant vector (black). At moderately low signal power, the addition of noise will effect the phase and rate of phase change. At very low signal power, the noise may cancel out the signal, causing the phase to change randomly, causing large rates of phase changes and the appearance of dropouts within the velocity signal.

### B.5.3 Signal noise generated by motions

If there is a lateral movement (lateral movement > beam diameter, such that the speckle pattern changes noticeably), a different speckle pattern arrives at the detector. There will be a high probability that a dark speckle passes over the detector at some time, which will cause a short period of low signal. This is known as speckle dropout.

Dropout reduction is possible by means of a tracking filter, which is used to bridge periods of low signal.
Appendix C

Electrocardiogram

C.1 Introduction

The LDI measurements setup is designed such that the measurements are synchronized with the heart rate frequency: the measurement of the next scanpoint in the sequence is triggered by the occurrence of a heartbeat. An electrocardiograph (ECG) is used to identify the QRS complex within the ECG waveform and to determine the moment at which the heart starts contracting. This appendix summarizes the most important aspects of electrocardiography (ECG) and describes two of the most common lead configurations, the classical Einthoven triangle and the widely used 12-lead ECG. A more detailed discussion of this topic can be found in Dupre et al. [2009].

C.2 The Electrocardiogram

An electrocardiogram (ECG) is used to measure the change in electrical activity during each cardiac cycle, and can be used, among other things, to identify the moment of heart contraction. Electrodes are placed around the heart in order to measure the potential differences along several directions and their change as the heart depolarizes and repolarizes.

During the cardiac cycle, the muscle cell membranes experience a small drop and rise in electric potential, the so-called action potential, and the muscle cells respond by contracting. During each cardiac cycle, action potentials move along the heart chambers and the heart contracts. In normal conditions, one part of the cardiac tissue is depolarized and other parts are at rest or polarized. This charge separation causes a current flow in the surrounding body fluids between the heart ends, resulting in fluctuating electric fields throughout the body. An ECG records these electrical fields by means of electrodes placed on the skin. The location of the electrodes is important, because the intensity of the voltage depends on the orientation of the electrodes with respect to the ends of the dipole.

It is important to note that an ECG is not a direct measure of the cellular depolarization and repolarization, but a measurement of the combined signals produced by the cell membrane potential changes of the surrounding tissue.

Furthermore, contractions of other muscles than the heart muscles, also result in observable action potentials (for instance measured in electromyograms) and will interfere with ECG recordings. In order to avoid this interference, it is common practice to measure ‘resting ECGs’ obtained in motionless patients.
C.3 ECG waveform

As mentioned above, electrode position can have a significant impact on the resulting ECG signal. However, irrelevant of the precise electrode position, some distinct artifacts can be identified, related to the ventricular or atrial depolarization and repolarization of the heart:

**P-wave** The depolarization of the both atria, begin of atrial contraction;

**QRS-complex** The depolarization of the left and right ventricles, after the S-wave the ventricles are depolarized and start to contract. The aortic valve opens and the aortic pressure begins to rise, resulting in the arterial systole. At the same time both atria repolarize, but this is not visible in the QRS-complex;

**T-wave** The repolarization of the ventricles.

Figure C.1 shows a comparison between the ECG waveform and other events in the cardiac cycle.

![Figure C.1: Comparison between the events displayed in an ECG compared with blood pressure within the aorta, ventrical pressure and volume, atrial pressure and the signal displayed in a phonocardiogram. It can be observed that the QRS wave is followed by the contraction of the ventricals.](image)

C.4 Lead configuration

Because the precise ECG waveform is dependent on electrode position, an ECG is usually recorded over multiple electrode positions, standardized by universal application and conventions.

The most commonly employed lead positions are referred as Lead I, Lead II and Lead III, as described by the Einthoven triangle (figure C.2). The vertices of the triangle can be considered to be at the wrists and left ankle, but also the shoulders and lower torso are adequate.

Another configuration is the 12-lead ECG, which the uses the three bipolar limb electrodes of the Einthoven triangle, 3 additional ‘uni-polar’ limb electrodes and 6 additional (uni-polar) chest electrodes, as shown in figure C.3.

The ‘Uni-polar’ limb leads use one limb electrode and a ‘neutral reference lead’, created by hooking up the other two limb locations to the negative lead of the ECG amplifier. The voltage recorded
between the left-arm limb lead and the neutral reference lead is called Lead aVL; similarly, the right-arm limb lead is aVR, and the left-leg lead is aVF (Fig. 17.11).

The remaining 6 of the 12 lead recordings are the chest leads. These leads are also unipolar; however, they uniquely measure electrical activity in the traverse plane instead of the frontal plane. Similar to the unipolar limb leads, a neutral reference lead is “created”, but this time using all three limb leads connected to the negative ECG lead, which basically puts it in the center of the chest. The six positive, or “exploring” electrodes are placed as shown in Fig. 17.12 (around the chest) and are labeled V1 through V6.

This configuration is widely used to evaluate multiple “views” of cardiac electrical activity, aiding in the localization of cardiac abnormalities.

Figure C.3: *Limb and chest lead configuration used in the widely used 12 lead ECG.*
Signal Analysis Theory

D.1 Introduction

This appendix is an attempt to give a summarized theoretical background of the signal analysis techniques employed within this thesis. It will treat the basics behind the Fourier Transform, Wavelet analysis and the generation of Frequency Response Functions. Although most of these concepts can be regarded as basic engineering knowledge, this appendix is added in order to present these concepts for readers without an engineering background.

D.2 Fourier Transform basics

D.2.1 Mathematical Definitions

Basis

A basis of a vector space \( \mathbf{V} \) is a set of linearly independent vectors, such that any vector \( \mathbf{v} \) in \( \mathbf{V} \) can be written as a linear combination of these basis vectors. There may be more than one basis for a vector space. However, all of them have the same number of vectors, and this number is known as the dimension of the vector space. Equation (D.1) shows an example of a vector \( \mathbf{v} \) written as a linear combination of the basis vectors \( b_k \) and the corresponding coefficients \( \nu^k \).

\[
\mathbf{v} = \sum_k \nu^k b_k
\]  

(D.1)

D.2.2 Basisfunctions

A similar concept can be generalized to functions, using basis functions \( \phi_k(t) \) to obtain a linear combination for function \( f(t) \) (equation (D.2)).

\[
f(t) = \sum_k \mu_k \phi_k(t)
\]  

(D.2)
The inner product of two basis functions is defined as:

\[ \langle f(t), g(t) \rangle = \int_a^b f(t) \cdot g^*(t) \, dt \] (D.3)

Two functions \( f \) and \( g \) are orthogonal to each other if their inner product is zero:

\[ \langle f(t), g(t) \rangle = \int_a^b f(t) \cdot g^*(t) \, dt = 0 \] (D.4)

A set of functions \( \phi_k(t), k = 1, 2, 3, \ldots \) is said to be orthonormal if they satisfy the conditions of equation (D.5):

\[ \int_a^b \phi_k(t) \cdot \phi^*_l(t) \, dt = \delta_{kl} \] (D.5)

With \( \delta_{kl} \) denoting the Kronecker delta function.

D.2.3 Definitions of the Fourier Transform

Continuous Fourier Transform

The Continuous Fourier Transforms (FT) decomposes a signal into its frequency components. The transform uses complex exponential (sines and cosines) functions as the basis functions. These basis functions are orthogonal, resulting in desirable properties which can be exploited during reconstruction. A drawback of the (continuous) Fourier Transform is its inability to contain time information: a time representation contains time resolution but no frequency information, while the Fourier Transform contains frequency resolution, but no time information. This limits usefulness of the Fourier Transform in the signal analysis of signals which contain localized properties, for example transients, or any signal of finite extent.

\[ X(f) = \int_{-\infty}^{\infty} x(t) \cdot e^{-2j\pi ft} \, dt \] (D.6)

In which:

- \( X(f) \) represents the Fourier Transform at frequency \( f \)
- \( x(t) \) denotes the time representation of the signal

Discrete Fourier Transform

In order to digitally process data, it is required to sample analog signals and convert them to digital data using an AD-converter. The Fourier Transform of the sampled data is defined by the Discrete Fourier Transform (DFT). In this case there is a finite number \( N \) samples of the signal \( p(t) \) taken at sample interval \( T_S \). After sampling for \( T = NT_S \), a set of \( \{ p_n \} \) of samples taken at regular intervals is obtained. It is possible to define

\[ p_n = p(t_n), \text{ where } t_n = nT_S, \text{ for } n = 0, \ldots, N - 1 \] is the sample coordinate. (D.7)
The DFT the transform is defined such that not only the input signal is defined at discrete points, but also the Fourier Transform is defined at regular points in the frequency domain: function $P(\omega)$ is only defined on frequencies $\omega_m$ instead of all frequencies. The samples $P(\omega_m)$ are defined to be regularly spaced at multiples of of the dominant frequency $\frac{1}{T}$, such that:

$$\omega_m = m \left( \frac{2\pi}{T} \right), \text{ for } m = 0, \ldots, N - 1 \quad (D.8)$$

It is assumed that the number of samples in frequency is equal to the number of samples in the temporal domain $N$. This condition is not necessary, but simplifies notation.

By inserting above definitions of $t_n$ and $\omega_n$ into the continuous FT (equation D.6), results in:

$$P(\omega_m) = \sum_{n=0}^{N-1} p(t_n) e^{-j\omega_m t_n} \quad (D.9)$$

Assuming that $\omega_n$ can only obtain discrete values (equation D.8) and $t_n$ can only assume discrete values as defined in equation D.7, the DFT (equation D.9) can be rewritten as:

$$P(\omega_m) = \sum_{n=0}^{N-1} p(t_n) e^{-j(m \frac{2\pi}{N})} e^{j(nT_s)} = \sum_{n=0}^{N-1} p(t_n) e^{-j\left( m \frac{2\pi}{N} \right) (nT_s)} \quad (D.10)$$

Simplifying and expressing the dependence on $\omega_m$ only in terms of $m$ and the dependency of $t_n$ only in terms of $n$ yields the final definition of the DFT:

$$P(m) = \sum_{n=0}^{N-1} p_n e^{-j \frac{2\pi}{N} mn} \quad (D.11)$$

And the inverse of the DFT (IDFT) is defined as [Fisher 2010, Keiner et al. 2009]:

$$p_n = \frac{1}{N} \sum_{m=0}^{N-1} P(m) e^{j \frac{2\pi}{N} mn} \quad (D.12)$$

### D.2.4 Non-uniform Discrete Fourier Transform

The need of analysis of irregularly sampled data arises due to recent developments in scientific disciplines like astrophysics, geoscience, seismics, remote sensing and medical imaging. A common approach is to resample the irregular data on a regular grid, although it is of interest to explore the possibility of analytical tools capable of dealing directly with irregularly sampled data.

For instance, in linear image processing, it is necessary to determine the convolution of input signal $p(t)$ with a filter $g(t)$, such that the output signal $y(t) = p(t) * g(t)$ highlights particular characteristics of the original input signal. The convolution can be simplified to a multiplication $Y(\omega) = P(\omega) G(\omega)$ using the Fourier transform. The investigation of the Fourier transform in the case of irregularly sampled input signal is therefore of great interest, since the knowledge of the Fourier transform in the irregular sampling case allows one to perform convolution in the irregular case and therefore opens the possibility of performing linear Image Processing on irregularly sampled signals.

In order to define a Non-uniform Discrete Fourier Transform (NDFT) is is necessary to generalize equation D.9 from regular sampling to irregular sampling. It is possible to apply irregular sampling in both the time $t_n$ and / or in frequency $\omega_m$, but most cases (like in this thesis) it is sufficient to consider the more restricted case of irregular sampling in the time domain but to use regular
sampling in the frequency domain: samples \( P(m) \) of the irregular FT are taken at multiples of \( \Delta k \), a fixed quantity in the Fourier domain. The fixed quantity \( \Delta k \) in the regular case is equal to \( \frac{2\pi}{T} \). The extension from regular to irregular sampling, therefore, depends on the duration of signal \( p(t) \) and not on the fact that the samples \( t_n \) are taken at regular or irregular intervals.

The definition of the non-uniform discrete Fourier transform (NDTF) becomes:

\[
P(m) = \sum_{n=0}^{N-1} p_n e^{-jm\Delta k t_n} \tag{D.13}
\]

Setting \( \Delta k = \frac{2\pi}{T} \), where \( T \) is the range of extension for the samples \( t_n \). In this case the NDFT is very similar to the DFT, except of the presence of the spatial coordinates \( t_n \) instead of index \( n \). Inserting gives the following definition of the NDTV [Fisher [2010]]:

\[
P(m) = \sum_{n=0}^{N-1} p_n e^{-j\frac{2\pi}{T}mt_n} \tag{D.14}
\]

Two major differences are to be observed between the DFT and the NDFT:

1. The samples in in frequency are taken at intervals \( \frac{2\pi}{N} \) in the irregular case instead of \( \frac{2\pi}{T} \) in the regular case (where \( T \) is the duration of the signal, \( N \) is the number of samples of the signal).

2. Instead of the integer index \( n \) in the regular case, the irregular sample coordinate \( t_n \) appears in the exponent.

It is possible to reconstruct the signal from the NDFT, but it should be realized that the definition of the iNDFT is not canonical (as observed in its matrix representation), and its matrix representation usually does not have an inverse. Instead, it is customary to define the adjoint NDFT [Keiner et al. [2009]].

### D.2.5 Short Time Fourier Transform

The Short Time Fourier Transform (STFT) is an alternative Fourier Transform developed in order to contain both time as frequency resolution. It is based on applying the Fourier Transform on local sections of the signal and yields the frequency content of the signal for each section. One of the downfalls is that the transform is defined with a fixed resolution: increasing the width of the section increases the number of samples, improving frequency resolution at the cost of a smaller number of sections available within the signal, reducing time resolution. Improving time resolution requires increasing the number of evaluation sections, reducing the amount of samples within each section, causing a reduction of frequency resolution. Using the STFT in an analysis requires a trade off between time and frequency resolution.

\[
STFT(X) (t', f) = \int x(t) \star \omega^* (t - t') \cdot e^{-j2\pi ft} dt \tag{D.15}
\]

In which:

- \( STFT(t', f) \) represents the short time Fourier transform for time \( t' \) and frequency \( f \).
- \( x(t) \) denotes the time representation of the signal.
- \( \omega(t) \) denotes the window function.
- \( \star \) denotes the complex conjugate.
D.3 Definitions of the Wavelet Transform

D.3.1 Continuous Wavelet Transform

Similar to the STFT, the Wavelet Transform (WT) provides a time-frequency representation. It is an alternative, developed to overcome some of the resolution problems occurring when applying STFT. The WT of a signal can be plotted on a 3-D graph, with time in one axis, frequency on the second and amplitude on the third axis. This plot reveals which frequencies exist at which time, similar to the SFTF, but with an improved resolution. In contrast to the fixed resolution used in STFT, the WT uses a variable resolution, dependent on frequency and time. Higher frequencies are better resolved in time, lower frequencies are better resolved in frequency.

In contrast to the Fourier Transform, which expresses the signal as a summation of (stationary) of sinusoids with distinct frequencies, the Wavelet Transform expresses the signal in the form of its similarity to (compressed and shifted) version of the mother wavelet [equation D.16][Polikar[2010]].

The CWT can be seen as the inner product of the test signal with the basis functions $\psi(\tau, s)(t)$

$$CWT_{\psi}^{x}(\tau, s) = \Psi_{\psi}^{x}(\tau, s) = \int x(t) \cdot \psi^{*}_{\tau, s}(t) \, dt$$ (D.16)

Where:

$$\psi^{\ast}_{\tau, s} = \frac{1}{\sqrt{s}} \psi \left( \frac{t - \tau}{s} \right)$$ (D.17)

In which:

- $\Psi(\tau, s)$ represents the continuous wavelet transform
- $\tau$ denotes the translation parameter:
- $s$ denotes the scale parameter: the inverse of frequency.
- $\psi(t)$ denotes the 'mother wavelet': the transforming function.
- $x(t)$ denotes the time representation of the signal.

Strictly speaking the scale $s$ used in wavelet analysis is related to the scaling / extension of the mother wavelet, a non-periodic, non-steady signal of finite duration. However, it is possible to associate the given wavelet to a purely sinusoidal signal of frequency $F_c$, $F_c$ being the frequency which maximizes the FFT of the wavelet modulus. This center frequency based approximation captures the main wavelet oscillations, leading in a simple characterization of the leading dominant frequency of the wavelet as a so called pseudo-frequency [Misiti et al. [2010]](figure D.1).

$$F_s = \frac{F_c}{s \cdot \Delta}$$ (D.18)

where:

$s$ is a scale;
$\Delta$ is the sampling period;
$F_c$ is the center frequency of a wavelet in Hz;
$F_s$ is the pseudo-frequency corresponding to scale $s$, in Hz.

Several possible mother wavelets can be used, eventually all the windows used in the analysis are the dilated (or compressed) and shifted versions of the mother wavelet.
Examples of mother wavelet functions

For example, the Mexican Hat wavelet is defined as:

$$\psi(t) = \frac{1}{\sqrt{2\pi\sigma^3}} \left( e^{-\frac{t^2}{2\sigma^2}} \cdot \left( \frac{t^2}{\sigma^2} - 1 \right) \right)$$  \hspace{1cm} (D.19)

Where:

- $\sigma$ denotes the scaling parameter that affects the width of the window

Figure D.1 shows some examples of other mother wavelets and their corresponding center-frequencies.

![Graphs of wavelets and center-frequencies](image)

**Figure D.1:** Examples of wavelets (blue) and their corresponding center-frequencies (red).

Although the wavelet itself are not periodic, its possible to associate a purely sinusoidal signal of center frequency $F_c$, maximizing the FFT of the wavelet modulus. This center frequency can be used to define the 'pseudo-frequency' associated with the wavelet and scale.

### D.3.2 Discrete Wavelet Transform

A discrete version of the Wavelet transform has been developed such that it can be approximated by numerical analysis. In contrast to the Discrete Fourier Transform, which simply samples the time-frequency (scale) plane with a uniform sampling rate, the scale change can be used to reduce the sampling rate when sampling for the Wavelet Transform.

According to Nyquist’s rule, the sampling rate can be decreased at higher scales (lower frequencies); if the time-scale plane needs to be sampled with a sampling rate $N_1$ at scale $s_1$, the same plane can be sampled with a sampling rate of $N_2$ at scale $s_2$, where $s_1 < s_2$ and $N_2 < N_1$. This relationship
Figure D.2: An 3-d example of a wavelet analysis, adjusted for pseudo-frequency. The Z-axis depicts the intensity of the WT. A logarithmic pseudo-frequency scale is used in order to represent the fine frequency resolution at low frequency (high scale) and course resolution at high frequencies (low scale). It can also be observed that time resolution increases at higher frequencies. These properties contrast with the STFT, which uses a fixed time and fixed frequency resolution.

satisfies:

\[ N_2 = \frac{s_1}{s_2} N_1 \]  

(D.20)

The continuous waveform (equation D.17) is discretized using the scale discretization \( s = s_0 \) and the translation discretization \( \tau = k.s_0 t_0 \), yielding the discretized Wavelet transform (equation D.21):

\[ \psi_{j,k}(t) = s_0^{-j/2} \psi\left(s_0^{-j}t - k\tau_0\right) \]  

(D.21)

It is also possible to define a Wavelet series transform (equation D.22), requiring \( \psi_{j,k}^*(t) \) to be orthonormal, biorthogonal, or frame.

\[ \psi_{x,j,k} = \int x(t) \psi_{j,k}^*(t) \, dt \]  

(D.22)

When \( \psi_{j,k} \) is not orthonormal, the wavelet series transform becomes:

\[ \psi_{x,j,k} = \int x(t) \psi_{j,k}^*(t) \, dt \]  

(D.23)

In which \( \hat{\psi}_{j,k}^*(t) \) is either a dual biorthogonal basis or a dual frame.

The discretized continuous wavelet and wavelet series are the result of applying sampling on the CWT, and as a consequence, require redundant information, requiring a significant amount of
computation time and resources. The Discrete Wavelet Transform (DWT) has been developed to yield sufficient information for both analysis as reconstruction, with a significant reduction in computation effort.

The procedure starts with passing the signal to a halfband digital low-pass filter $h[n]$. The convolution operation in discrete time is defined as:

$$x[n] * h[n] = \sum_{k=-\infty}^{\infty} x[k] \cdot h[n-k] \quad (D.24)$$

### D.4 Frequency Response Function

#### D.4.1 Auto Power Spectrum

The frequency content of the signal is evaluated using the Auto Power Spectrum (APS). The APS (equation D.26) is calculated using the (Fast) Fourier Transform, as implemented in the Matlab function \texttt{fft}. Additional scaling is necessary on the result (shown in equation D.25) to compensate for:

(a) Multiply each component (except the DC component corresponding with 0 Hz) with 2 compensate because the Matlab function \texttt{fft} yields the two-sided FFT of $x(t)$.

(b) Divide by N in order to compensate for sample length of $x(t)$.

(c) NFFT: the amount of elements used in FFT: if NFFT exceeds the sample length N, the signal $x(t)$ will be padded with trailing zeros to length NFFT.

(d) Application of a window function $w(t)$ in order to reduce spectral leakage in the frequency spectra

$$X_{FFT}(f) = \frac{2 \cdot FFT(w(t) \cdot x(t),NFFT)}{N} \quad (D.25)$$

$$APS_{xx} = X_{FFT}(f) \cdot \overline{X_{FFT}(f)} \quad (D.26)$$

When averaging auto- and crosspower spectrums (in the frequency domain) one should be aware of the fact that they are statistically unique only if the input is stationary and sufficient averages have been taken.

#### D.4.2 Frequency Response Function

It is possible to determine a transfer function estimate between two signals by defining the Frequency Response Function $H(f)$, defined in equation (D.27). Depending on the to be expected noise origins, there are three different FRF estimates defined $H_1$, $H_2$ and $H_v$. The first defines the best estimate in the case where it is assumed that there is no noise on the input and noise is only present on the response, the second defines the best estimate in the case where the noise is assumed to be present on the input and there is no noise on the output signal. The latter gives the best estimate in the case where the noise is assumed to be both present on the input as well as on the output signal.
Figure D.3: Auto Power Spectrums of $\frac{dP}{dt}$, obtained by several different windows functions.

The blue spectrum shows the APS when applying a rectangular window, the green spectrum is the result of a Hann window, the red spectrum corresponds with a Blackman-Harris window and the magenta spectrum is the result of applying an exponential function with $\tau = 2.5$.

As showed in literature, the cardiac cycle will most definitely not be the only input for abdomen motion. Respiration, and maybe the presence of peristaltic motions, will cause the abdomen surface to show motion not origination from the cardiac input.

A practical way to calculate the FRF is presented in equation (D.28), the quotient of the Cross Power Spectral Density (CSD) ($CPS_{xy}$) of $x$ and $y$ and the Power Spectral Density $APS_{xx}$ of $x$.

$$X(f) = H(f) F(f) \quad \text{(D.27)}$$

$$H_1(f) = \frac{CPS_{xy}}{APS_{xx}} \quad \text{(D.28)}$$

The process of signal averaging as applied to frequency response functions is simplified greatly by the uniqueness of the frequency response function. This concept means that when formulating the frequency response function using cross- and autopower spectrums, the estimate of the frequency response function is intrinsically unique as long as the system is linear. This estimate is valid whether the input is stationary, non-stationary or deterministic.

In case of asynchronous signals $H(f)$ must be calculated by averaging the auto- and crosspower spectrums, since no other way of preserving phase and improving the estimate is available. Only in case of synchronous signal averaging the formulation, but this is rarely done [Heylen et al. 1998].

In practice the frequency response function will be estimated by averaged values (indicated by $\hat{H}$)
Figure D.4: Auto Power Spectrums of the velocity signal, using several different window functions. The rectangular window (blue) shows severe spectral leakage. The Tukey ($r = 0.5$) window (cyan) is essentially a hybrid of a rectangular and a Hann window, and also shows signs of spectral leakage. The Hann (green), Blackman-Harris (red) and Flat-Top (magenta) window functions appear to be show quite similar results, although Flat-Top appears to be lacking some detail as it does not show a peak at 7 Hz like the other window functions.

of the auto- and crosspower spectrums (equation D.29) [Schwarz and Richards 1999].

\[
\hat{APS}_{xx} = \frac{1}{N_a} \sum_{n=1}^{N_a} (APS_{xx})_n, \quad \hat{APS}_{yy} = \frac{1}{N_a} \sum_{n=1}^{N_a} (APS_{yy})_n, \quad \hat{CPS}_{yx} = \frac{1}{N_a} \sum_{n=1}^{N_a} (CPS_{yx})_n \quad (D.29)
\]

As previously mentioned, the choice of windowfunction influences both APS and CPS. The influence of windowfunction on the FRF is shown in figure D.5(a).

D.4.3 Coherence function

The Coherence between two functions indicates how well signal $x$ corresponds to signal $y$ at each frequency and ranges between 0 and 1. It is a function of the PSD’s $P_{xx}$ and $P_{yy}$ and the CSD $P_{xy}$ between the two signals, as defined in relation D.30. As figure D.5(b) reveals, coherence is effected by the choice of windowfunction.

\[
C_{xy} = \frac{\hat{CPS}_{yx}^2}{\hat{APS}_{yy} \cdot \hat{APS}_{xx}} \quad (D.30)
\]
Due to the small sample length of the signals, quite a lot of information is lost when applying a window function. However, as will be shown, spectral leakage is an important issue (especially when applying ‘zero-padding’ to increase frequency resolution and focusing the analysis on the [0 - 50 Hz] bandwidth). It is possible to reduce spectral leakage by implementing a window function, but it should be realized that the sample length of the signals is quite small. As previously mentioned, most power is centered in the 0 to 20 Hz band. Each sample only contains a few instances of the input, and the most interesting feature, the so called ‘cardiac sine’ occurs at the beginning of the sample. As shown in figure D.5(b), the application of a window function would have significant effect on these features, making appropriate window choice an important issue within
Due to the small sample periods, the application of windowing will have a profound effect on the signals. It is interesting to note that the ref-signal, corresponding with the blood pressure, does not have an average zero power; This non-zero DC power will be present as a large DC frequency component within the frequency analysis.

D.5.2 Resolving spectral leakage by windowing

Measuring the signal over a finite time is essentially equivalent to a multiplication of the (infinite) signal with a rectangular function of unit amplitude lasting over the measurement time. The convolution (in the frequency domain) effectively results in the multiplication of each separate (sampled) frequency with a ‘sinc’ function or the so-called ‘Dirichlet kernel’. This convolution causes the real frequency component to leak e.g. smear over several ‘frequency bins’ of the Dirichlet kernel (figure D.7).

It is possible to reduce the effects of the discontinuities by multiplying the signal with a ‘window function’ which smoothly reduces to zero at the end points, and is able to reshape the amount of spectral leakage occurring between the frequency bins to more favorable magnitudes.

Window properties

Choosing an appropriate window function is not arbitrary; applying a window function is the result of a trade off between noise reduction and frequency resolution. Several window parameters have been defined to evaluate the performance of a window function:

**Equivalent Noise Bandwidth (ENB):** The ENB of the window measures the noise performance of the window. It is the width of the rectangular filter which would accumulate the same noise power with the same peak power gain.
Figure D.7: Example of a rectangular window (width $2\pi$) and its Fourier Transform. Sampling over a finite time is essentially equivalent to a multiplication with the (infinite) signal of a rectangular function with a unit amplitude lasting over the measurement time (as in figure D.7(a) over a sample time of $-\pi$ to $+\pi$). The convolution (in the frequency domain) with the Dirichlet Kernel (figure D.7(b)) causes the separate frequency contribution to be smeared over multiple frequency bins, hence spectral leakage.

Coherent Power Gain: The Coherent Power Gain measures the reduction in signal power due to the window function suppressing a coherent signal at the ends of the measurement interval.

Worst case processing loss: The worst case processing loss is defined as sum of Scalloping loss and Processing loss:

- **Processing loss**: Processing loss is the ratio of input signal to noise ratio to output signal to noise, which is the Coherent Power Gain divided by the noise power.
- **Scalloping loss**: Scalloping loss is defined as the ratio of coherent gain for a signal frequency component located half way between FFT bins, to the coherent gain for a signal frequency component located exactly at an FFT bin.

The worst case processing loss is a measure of the reduction of output signal to noise ratio resulting from the combination of the window function and the worst case frequency location.

Sidelobe level - Fall off: The peak sidelobe level indicates how well a window function suppresses spectral leakage, so is the fall off to the sidelobes. The larger the reduction, the less does the frequency component at the peak ‘leak’ to the nearby peak sidelobes or other sidelobes.

6 dB bandwidth: indication of minimum resolution? minimum separation of two freq. comp at equal amplitude, so they can be resolved. (a local minimum is present between the two). If two freq. components are involved in the sum of the window functions at the cross over point (halfway between the peaks) must be smaller than the individual peaks if the two peaks are to be resolved. so the gain from each window function must be less than 0.5 (or 6 dB).

D.5.3 Examples of window functions

Several different windows function have been defined, which with the above described properties, are suited for particular situations.
Rectangular window

The rectangular window is highly suitable

- for signals which are periodic in the window,
- or non-periodic signals which decrease to zero within the time window (e.g. hammer blows).

The rectangular window is not suitable

- for higher and very high frequency resolutions, as the amplitude attenuation of the highest secondary maxima is low in the spectrum (highest side lobe attenuation is low) and the amplitudes of the secondary maxima only decrease slowly (roll-off rate is low).

The rectangular window does not falsify the spectrum of signals which are periodic in the time window. The rectangular window strongly falsifies the spectrum of signals which are non-periodic in the time window.

Hanning window

The Hanning window has the shape of an inverse, lifted cosine function. With noisy measurement signals occurring in a noise excitation, a window function should display even-sized attenuating behavior in the complete window spectrum.

The Hanning window is highly suitable for strong noisy measurement signals as

- the amplitudes of the secondary maxima decrease rapidly (roll-off rate is high),
- the main maximum is only double the width compared to that in the rectangular window spectrum,
- the attenuation is constant.

periodic in the time window more than the rectangular window, non-periodic in the time window less than the rectangular window.

Hamming window

The Hamming window is a combination of the Hanning window and the rectangular window.

In comparison to the Hanning window, the Hamming window is more suitable

- for the frequency resolution of spectral lines in attenuations above the highest side lobe attenuation.

The Hamming window is not better suited than the Hanning window

- for very noisy measurement signals and the frequency resolution of spectral lines in an attenuation beneath the highest side lobe attenuation of the Hamming window, as the amplitudes of the secondary maxima only decrease slowly (roll-off rate is low) and the main maximum is double the width as compared to the rectangular window spectrum.
<table>
<thead>
<tr>
<th>Window function:</th>
<th>Rectangular</th>
<th>Hann</th>
<th>Hamming</th>
<th>Blackman-Harris (4-p)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sidelobe level (dB)</td>
<td>-13</td>
<td>-32</td>
<td>-43</td>
<td>-92</td>
</tr>
<tr>
<td>Fall off (dB per oct)</td>
<td>-6</td>
<td>-18</td>
<td>-6</td>
<td>-6</td>
</tr>
<tr>
<td>Coherent gain</td>
<td>1.00</td>
<td>0.50</td>
<td>0.54</td>
<td>0.36</td>
</tr>
<tr>
<td>Eq. noise Bandw. (bins)</td>
<td>1.00</td>
<td>1.50</td>
<td>1.36</td>
<td>2.0</td>
</tr>
<tr>
<td>6 dB Bandw. (bins)</td>
<td>1.21</td>
<td>2.00</td>
<td>1.81</td>
<td>2.72</td>
</tr>
<tr>
<td>Worst case proc. loss (dB)</td>
<td>3.92</td>
<td>3.18</td>
<td>3.10</td>
<td>3.85</td>
</tr>
</tbody>
</table>

**Table D.1:** Performance parameters of several window functions. The type of window function is a trade off between the reduction in spectral leakage the window function can obtain and the signal loss caused by the application of a window function. The (4 term) Blackmann-Harris window has got a very strong sidelobe reduction, but at the expense of a wider main-lobe, reducing the maximum attainable frequency resolution. Parameters adapted from [Processing 2010] & [Lessard 2006].

**Blackman-Harris Window**

The Blackman-Harris window is part of the generalized Hamming family, and is optimized for minimum side-lobe (4 term level = $-96\,dB$) level and maximum roll-off rate (4 term roll-off rate = $-6\,dB$), making it very effective in suppressing spectral leakage. However, the reduction of side-lobe level comes at the cost of an increase in main lobe width (6 dB Bandw. (bins) = 2.72 bins), thereby reducing the frequency resolution (Table D.1).
Reliability of surface interpolation

E.1 Introduction

On several occasions within this thesis, plots are shown of the abdominal surface, for instance the Instantaneous Deflection Shapes shown in section 5.4.1, the plots showing Operation Deflection Shapes (section 5.6.3) and the surfaces used during the definition of the 2D spatial Fourier Transform (section 5.6.5). However, measurements where only obtained on a limited amount of gridpoints, not sufficient to able to plot a surface, and MATLAB interpolation algorithm was used to obtain data for a larger grid. This appendix will give more detail into use of this interpolation technique and will try to determine how reliable this interpolation really is.

E.2 MATLAB implementation

The surfaces are generated by using the MATLAB function `meshgrid` which uses a vector of Z-values \( z \) with coordinates \( (x, y) \) to interpolate values \( Z_I \) on the predefined grid \( (X_I, Y_I) \) (equation E.1):

\[
[X_I, Y_I, Z_I] = \text{griddata}(x, y, z, X_I, Y_I) \tag{E.1}
\]

The function `griddata` can use different interpolation methods, which can be enabled in the options:

- **linear** Triangle-based linear interpolation (default).
- **cubic** Triangle-based cubic interpolation.
- **nearest** Nearest neighbor interpolation.
- **v4** MATLAB 4 griddata method

E.3 Linear vs cubic interpolation

Figure E.1 shows a comparison between a linear and a cubic interpolated surface. It should be noted that, in contrast to previously shown surfaces, the faceplot has been set on flat. Most other
figures where usually plotted with faceplot set on interpolation, in order to obtain less checkered coloring. Using flat facecolors results in a more accurate representation of the calculated $Z_i$ values, making it easier to compare between interpolation results. Comparison between both methods does not show very large differences (figure E.1), so it appears that both methods give similar graphical results.

**Figure E.1:** Comparison of linear interpolated surface and a cubic interpolated surface

The order to compare both methods more thoroughly, the next step is to evaluate the difference between both interpolation methods, using the following error definition (eq. E.2):

$$
\varepsilon = Z_{I}(X_{I,i,j}, Y_{I,i,j})_{cubic} - Z_{I}(X_{I,i,j}, Y_{I,i,j})_{linear}
$$

(E.2)

Plotting for one particular time instance yields the following figure:

**Figure E.2:** Error $\varepsilon$ as defined in equation E.2 for a complete surface. It can be observed that error increases near gridpoints $(x, y)$, but becomes zero at the gridpoints itself. This is caused by the piece-wise construction of the linear approximation of the surface (figure E.3)

Figure E.2 shows that absolute error is at average between $-2E - 3m/s$ and $2E - 3m/s$, with maximum peaks reaching up to $\pm 6E - 3m/s$. The increase of error appear to coincide with the
gridpoints, which can be explained by the fact that the linear approximation constructs the surface in a piece-wise matter, while the cubic approximation creates a smooth surface. As a result the grad of the surface tends to zero in the cubic approximation, while the linear does not. This effect is demonstrated in figure E.3.

![Figure E.3](image)

**Figure E.3:** Figure E.1 showed that the error between both methods increased near peaks, but reduces to zero at the position of the gridpoints. This phenomenon is caused by a difference in surface gradients. The cubic approximation (red) tries to obtain a zero gradient at the gridpoints (grey), while the linear approximation (green) is just a piece-wise surface. Near peaks the difference $\varepsilon$ becomes quite large, while both approximations do pass through the same gridpoints.

### E.4 Number of interpolation points

Another interesting question is to evaluate the influence of the extend of interpolation resolution on the difference. In order to investigate the influence, a grid of $n = 50$ elements is compared to finer grids of $n^\alpha$, satisfying condition:

$$n^\alpha = n + \alpha (n - 1) \quad \text{(E.3)}$$

With $\alpha = 0, 1, 2, 3, \ldots$, corresponding with the amount of midpoints between the original points. When expanding to a 2-d case, equation E.4 defines the error estimate, and compares the absolute difference between the base mesh and the finer mesh,

$$\varepsilon^\alpha (k, l) = \left| ZI^{\alpha=0} (k, l) - ZI^\alpha (k + \alpha (k - 1), l + \alpha (l - 1)) \right| \quad \text{(E.4)}$$

Calculating this error for a real dataset (‘S6V6.mat’), and using a cubic interpolation method, yields the results shown in table E.1. The error is in the same order as the numerical accuracy used in the calculations. Experimenting with linear interpolation and even smaller grids yield the same conclusion. A finer interpolation resolution does not result in significant different estimations when compared with the results of a courser grid.

Experimenting with other datasets (table E.2) does not yield other results, the influence of increased gridnumber $n$ does not influence the accuracy of the estimation at fixed coordinates. It can be concluded that the the number of interpolation points does not influence the interpolation, but rather increases the number of samples on the interpolation.
## Table E.1: Calculation of the increase of interpolation accuracy when increasing the number of interpolation points, performed on the surfaces at three different time instances \( t_a, t_b, t_c \).

Calculating the error defined in Equation E.4 shows that an increase of resolution does not yield any improvement. The listed errors are in the same order as the numerical accuracy of MATLAB (double precision): \( \varepsilon = 2.2204e-16 \).

<table>
<thead>
<tr>
<th>alpha</th>
<th># of interpolation points n</th>
<th>'S7VS11.mat'</th>
<th>'S7VS3.mat'</th>
<th>'LUMC1_P1_A_V2.mat'</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>50</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>1</td>
<td>99</td>
<td>-</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>2</td>
<td>148</td>
<td>1.6989e-016</td>
<td>1.0712e-016</td>
<td>2.0231e-016</td>
</tr>
<tr>
<td>3</td>
<td>197</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>4</td>
<td>246</td>
<td>1.6989e-016</td>
<td>6.7112e-017</td>
<td>1.3585e-016</td>
</tr>
</tbody>
</table>

## Table E.2: An extension of the error calculations presented in table E.1. All the datasets show a similar behaviour. Increasing the resolution of the interpolation grid does result in an increase of estimation accuracy.

### E.5 Resolution error

The previous sections addressed errors involving the interpolation method. However, more interesting is the question if the finite amount of measured gridpoints is sufficient to characterize the motion of the surface, or that it may be possible that another wave is present between two different gridpoints.

This is a hard question to answer; the velocity patterns that where measured showed that the vibration frequency was generally in the order of 0 - 25 Hz, and generally do show some similar shape when comparing its velocity profile with the profiles of the surrounding gridpoints (as previously observed in plots like figure E.3). This suggests that, while it is necessary to distinguish spatial position, there are no sudden local changes in vibration type.

In order to check if the spatial resolution of gridpoint is very important for IDS interpolation, two different IDS plots have been defined (figure E.4): one using a full grid, the other by arbitrarily removing half of the points:

### E.6 Conclusion

Interpolation appears does not appear to influence the general shape. Both linear and cubic interpolations give a similar solutions, although the linear solution gives a cruder one. Increasing the gridsize does not appear to change the estimate at distinct coordinates, but it does increase the resolution.

Influence of gridpoint resolution was also tested. Even when reducing the number of evaluated gridpoints to its half, the interpolated surface still showed a similar general profile. However, it should be realized that reducing the gridpoint resolution, will result in loss of spatial details.
Figure E.4: Comparison of Instantaneous Deflection Surfaces of the averaged velocity profiles previously shown in figure 6.3 generated with a full grid of measurement locations and one with half of the gridpoints removed. Comparison between the full grid in subfigure E.4(a) with the half grid in subfigure E.4(b) shows quite some similarity. Although spatial details are lost, the general profile did not change much.