Visualization and Heuristic Modeling for Planning of Minimally- and Non-Invasive Tissue Ablation

by

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Statutory Declaration
(Declaration on Authorship of a Dissertation)

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Furthermore I declare that I have written this PhD thesis independently, unless where clearly stated otherwise. I have used only the sources, the data and the support that I have clearly mentioned.

This PhD thesis has not been submitted for the conferral of a degree elsewhere.

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Place                                      Date

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Signature
To Kathi, Lotta, Milla and Emma
Abstract

MINIMALLY- AND non-invasive tissue ablation procedures allow for the destruction of pathological tissue - such as tumors - deep within the human body. Needle-shaped instruments or high-intensity focused ultrasound are utilized to create a necrotic zone in the target area. These local approaches result in minimal or even no trauma for the patient and have become alternative treatments, especially for patients who are not surgical candidates. The identification of suited treatment strategies based on pre-interventional image data is necessary because the anatomical situation is not directly visible during intervention. A route to the target has to be identified which spares non-target structures. Furthermore, the position and extent of the necrotic zone resulting from the therapy has to be predicted and adapted to facilitate complete ablation of the target volume.

This PhD thesis describes methods for the support of the planning process for minimally- and non-invasive tissue ablation procedures. It focuses on the utilization of visualization and heuristic modeling to solve complex problems such as access path determination for needle-based tumor ablation therapies or sonication planning for high-intensity focused ultrasound.

For needle-based interventions, the manual selection of suited access paths is addressed, which typically requires the careful inspection of every candidate path as a whole to make sure that no structure at risk would be penetrated. Especially for angulated paths that are not included in one image slice, this results in repeated inspection of multiple slices. To improve this process, a visualization method for the highlighting of infeasible paths in common 2D viewers is proposed.

As an alternative to manual planning of percutaneous procedures, heuristic approaches to therapy plan optimization are investigated. An algorithm based on projection utilizing the GPU and image processing is proposed. Under consideration of multiple clinically relevant criteria, the method generates a list of optimal paths within a few seconds. A second, semi-automatic method is proposed, which represents a compromise between fully automatic and manual planning. It combines the fast projection based method with a numerical approach for Pareto-front approximation and allows for interactive exploration of the solution space.

To improve the manual planning of high-intensity focused ultrasound therapies, a real-time approximation of temperature and thermal dose is developed. It is based on numerical simulation for a range of exemplary configurations in a preprocessing step. During interactive planning, the temperature or thermal dose field for the situation at hand is interpolated and combined with an approximation of the heat sink effect resulting from vessels in proximity. The result is visualized in an interactive manner.
Acknowledgment

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1 Introduction

If I could time travel into the future, my first port of call would be the point where medical technology is at its best because, like most people on this planet, I have this aversion to dying.

(Neil Asher, Science Fiction Author)

Image-guided, minimally- and non-invasive interventions are an integral part of today’s radiological practice. These types of interventions result in minimal trauma for the patient and therefore reduce the efforts and costs inflicted by post-interventional care. As a subspecialty, tissue ablation therapies aim at the complete destruction of a target volume (e.g., a tumor) with a minimum amount of affected healthy tissue and without damaging adjacent vital structures. The planning of such procedures requires a high amount of experience and has to be carried out in an effective and safe manner in the clinical practice. It is the aim of this PhD thesis to investigate computer support for this complicated process. For this purpose, several methods based on visualization as well as heuristic modeling for therapy plan definition are investigated.
1.1 Medical Background

Among noncommunicable diseases, cancer is a leading cause of death [52, p. 24]. An estimated 14.1 million new cancer cases and 8.2 million deaths resulting from cancer occurred in 2012 worldwide [73]. Over 21.6 million new cases are expected for the year 2030 [52, p. 31]. Cancer of the lung, breast, colorectum, prostate, stomach, and liver have the highest incidence rates. The mortality is highest for cancer in the lung, liver, stomach, colorectum, breast, and prostate [73] (see Table 1.1). For both, lung and liver cancer, the standard of care procedure with the best curative perspective is surgical resection. However, fewer than 25% of lung cancer patients [70] and 15% of liver cancer patients [44] are candidates for surgery due to anatomical or patient co-morbidity factors. In other cases, such as prostate cancer, resection may be related to strong negative effects on the quality of life [189].

Over the last decades various image-guided minimally- and non-invasive tissue ablation procedures have been developed and established as an alternative to surgical resection. Ablation is the direct application of destructive processes in the target area, e.g. the tumor. These processes use, among others, heat or chemical agents as the means for tissue destruction, which are released or generated directly in the target area. The goal is the localized, complete destruction of pathological tissue in that region with a minimum amount of affected healthy tissue and without damaging adjacent vital structures. Theses methods are referred to being image-guided because medical imaging is utilized during all stages of the therapy.

Percutaneous ablation procedures such as radiofrequency ablation (RFA) or microwave ablation (MWA) utilize needle-shaped instruments to deliver the means of tissue ablation to the target site. Hence, the destructive effect, e.g. resistive heating in the case of RFA, originates from a portion of the instrument close to the tip. Since the placement of the instrument into the body results in small incisions only, percutaneous ablation modalities are also referred to as being minimally invasive. In contrast, high-intensity focused ultrasound (HIFU) requires no incision at all and destroys the target structure by focusing ultrasound beams onto it. Hence, it is often described as being non-invasive. As a result of the minimally- or even non-invasive

<table>
<thead>
<tr>
<th>Organ</th>
<th>Incidence</th>
<th>Mortality</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung</td>
<td>1.800.000</td>
<td>1.600.000</td>
</tr>
<tr>
<td>Breast</td>
<td>1.700.000</td>
<td>522.000</td>
</tr>
<tr>
<td>Colorectum</td>
<td>1.400.000</td>
<td>694.000</td>
</tr>
<tr>
<td>Prostate</td>
<td>1.100.000</td>
<td>307.000</td>
</tr>
<tr>
<td>Stomach</td>
<td>951.000</td>
<td>723.000</td>
</tr>
<tr>
<td>Liver</td>
<td>782.000</td>
<td>745.000</td>
</tr>
</tbody>
</table>

Table 1.1: Worldwide incidence and mortality estimates for the 6 most common cancers. Source: Ferlay et al. [73]
nature of tissue ablation procedures, overall costs are reduced and hospitalization times are shortened [57].
Tissue ablation procedures are indicated for both, *curative* (complete eradication of all known tumor cells within the treated tumor and without any other known tumor in the body) and *palliative* (complete ablation of the target tumor with other known tumors in the body or complete or partial ablation to achieve symptom relief) treatments [32] and have been used to treat cancer in the liver, kidney, lung, breast, prostate, and bone [65, 113, 57]. Among these, the liver has been the most studied target organ for ablation of both primary and metastatic malignancies [110]. Non-oncological applications include for example the treatment of prostatic hyperplasia, uterine fibroids, habitual snoring [65], and neurological diseases (essential tremor, Parkinson’s disease, depression [74, p.4]). A high potential for further applications results from combination with other therapies. The achievable size of the ablated volume can be increased by combination with radiation- or chemotherapy [33, 57]. Furthermore, the utilization of ablation procedures to enable targeted drug delivery [33, 113] or gene therapy [113] as well as synergies with immunotherapy [57] are current fields of research.

### 1.2 Challenges and Contributions

A major challenge of tissue ablation procedures is the planning of the intervention based on patient image data. For both, minimally- and non-invasive tissue ablation procedures one or multiple ablations have to be planned such that the target volume, or a defined subvolume of it, is completely destroyed. However, incomplete ablation and disease recurrence are common for ablation procedures. Although the results vary substantially between the modalities, the outcomes are generally inferior when compared to surgery [57]. Major reasons for incomplete ablation are the generally limited size of the achievable ablated volume due to technological constraints as well as the reduction of the ablated volume due to the perfusion of nearby vessels and capillary level microperfusion [86, 203]. On the other hand, the therapeutic effect may occur at unforeseen locations, leading to unwanted ablation of healthy tissue [185, 158, 67, 138, 37, 170, 77, 110]. Furthermore, in the case of percutaneous procedures, the placement of the instrument may lead to bleeding, pneumothorax, hemothorax, infections, and tumor seeding in the needle tract [158, 138, 77, 110]. Hence, the definition of a safe access path during the planning phase is essential [202, 138]. Planning and implementation of minimally- and non-invasive ablation procedures under consideration of the above-mentioned aspects require considerable experience. For example, a study evaluating the effect of operator experience on the treatment outcome for RFA of malignant liver tumors indicates that complete ablation and low complication rates can be achieved with the accumulated experience gathered during 50 interventions by a specialized team [150]. Strong variability in 5-year patient survival (14% to 55%) and local tumor recurrence rates (3.6% to
Chapter 1. Introduction

60\% in RFA are partially attributed to inexperience with the technique [203]. A similar influence of operator experience on outcomes seems to be associated with HIFU [45].

The aim of this work is the development of methods that can support the complicated planning process. Computer-assistance is able to reduce the complexity of the planning task, and therefore the planning duration, while raising safety and improving the outcome. While this especially applies to complicated cases, less experienced physicians might also benefit from computer-assistance even if the case at hand is not overly complicated. However, the restricted amount of time also limits the running time of any utilized algorithm. The time available for planning is not standardized and indeed a subject for debate. Some clinical workflows might allow for planning on the day before the intervention. However, planning directly before the intervention after patient preparation is more common. Hence, the time window available for planning is limited. Any approach of computer-assisted planning should consider this restriction to be of clinical relevance. Furthermore, the methods must be easy to use. Complex parametrization of algorithms should be avoided.

Traditionally, the planning of percutaneous ablation procedures is performed based on 2D slices of computed tomography (CT) or magnetic resonance imaging (MRI) scans. However, various criteria have to be taken into account during planning. Among other aspects, the consideration of the patient anatomy is crucial. The selected access path must not penetrate any structure at risk. Using the classic visualization alone, it can be challenging and time consuming to perform that planning task with the necessary care, in particular if no access path can be found that is contained in a single image slice. Hence, in this work two visualization methods to accentuate unsuited access paths are proposed. In contrast to other methods from this field, which mainly focus on the visualization of the patient anatomy in 3D visualizations, the presented methods can be seamlessly integrated into the established workflow based on the examination of 2D image slices. The first method utilizes the graphics processing unit (GPU) to create the risk structure map, a representation of the considered risk structures that allows for interactive highlighting of unsuited skin entry points. This basic method is extended to an advanced visualization method which, among other refinements, also allows for definition of the target point and for the consideration of collisions with other instruments in a multi-instrument setup. A clinical evaluation indicates that the proposed methods are suited to improve the time needed for planning as well as the safety of the defined access paths.

As an alternative to the manual definition of access paths, two optimization approaches are presented. In addition to the avoidance of risk structure penetration various clinically relevant criteria are considered. The first method is fully automatic and computes a list of optimal access paths for a given target point within a few seconds. This is achieved by a brute-force optimization method based on GPU-computed projections and image processing. The second method is semi-automatic and combines the first approach with a numerical optimization scheme which allows to integrate a fast and realistic GPU-based approximation of the ablation zone including cooling effects from nearby blood vessels. Interactive exploration of the
optimization results is facilitated by a slider-based graphical user interface (GUI). The results of two clinical studies for the evaluation of the two proposed methods are reported.

For the planning of non-invasive HIFU, a new interactive method for the approximation of the ablation outcome is proposed. The planning of HIFU ablations is typically concerned with the placement of multiple ablation zones, as the achievable single ablation zones are very small. The common approach in the literature to support this planning task involves computationally expensive numerical simulations [62, 97, 128, 207, 4, 210]. The proposed method represents a surrogate that can be used for initial interactive planning, before a more realistic simulation is carried out. To achieve an interactive visualization, a multi-dimensional lookup table is created based on numerical simulation. It contains compact, interpolatable approximations of the ablation outcomes for many parameter combinations. During visualization, the outcome for a certain situation is approximated by interpolation. This approach is combined with an approximation of the cooling effect of nearby vessels, which is also based on numerical simulations.

1.3 Structure of this Thesis

This thesis is structured as follows. After a description of the medical background and technical fundamentals relevant to this work (Chapter 2), novel techniques for the planning of minimally- and non-invasive interventions are introduced:

- **Chapter 3** introduces interactive techniques for the visualization of infeasible access paths for the planning of percutaneous procedures. In order to support the selection of safe access paths, the presented methods address the identification and visualization of paths that would penetrate previously segmented risk structures.

- **Chapter 4** describes new methods for automatic and semi-automatic selection of optimal access paths for percutaneous tumor ablation. The proposed optimization approaches incorporate multiple clinically relevant criteria in order to create access path proposals of clinical value.

- **Chapter 5** presents a novel interactive approach for the prediction of the ablation outcome for HIFU planning. The method approximates the result of a numerical simulation including a realistic estimation of the cooling effects resulting from nearby vessels.

Finally, **Chapter 6** provides the conclusion of this PhD thesis with respect to the objectives of this thesis and outlines possible future research directions.
2 Fundamentals and State of the Art

Any sufficiently advanced technology is indistinguishable from magic.

(Sir Arthur C. Clarke)

This chapter explains the principles of image-guided minimally- and non-invasive tissue ablation procedures and describes the most important ablation modalities. Furthermore, considerations taken into account during the planning of the procedures are summarized and the established planning workflows are described. Relevant concepts and methods from the fields of computer graphics and optimization are introduced.
Chapter 2. Fundamentals and State of the Art

2.1 Image-Guided Tissue Ablation Procedures

Over the last decades, various modalities for tissue ablation utilizing different physical principles have been developed. In the following, common terminology used in this work is summarized and the typical procedure scenario (Section 2.1.1) and treatment goals (Section 2.1.2) are described. Section 2.1.3 outlines the classification of tissue ablation procedures. After a description of the most commonly used modalities (Section 2.1.4), the role of image-guidance as well as the main stages of tissue ablation procedures are discussed in Section 2.1.5.

2.1.1 Common Terminology and Procedure Scenario

Radiofrequency ablation (RFA), one of the first broadly used tissue ablation modalities, was pioneered by radiologists in the 1990s [135, 166]. Today, tissue ablation procedures are mainly performed by interventional radiologists [142, 176, 104, 133] and interventional oncologists [179] in many countries, but may also be performed by gastroenterologists and internists [176]. Furthermore, ablation modalities are used by surgeons in open surgical procedures [90]. For the remainder of this work, the term physician will be used to refer to the person planning and performing the procedure in a general manner.

The overall ablation scenario contains the ablation device, which delivers the ablation effect [168] to the target site and an additional device close to the treatment site, to which the ablation device is connected. These so-called generators [46, 69, 32, 63] or consoles [131] provide the means of tissue destruction and control the applied dose. For the remainder of this work the more general term generator is used. An imaging device, typically computed tomography (CT), magnetic resonance imaging (MRI) or ultrasound (US), is utilized to support planning, implementation and control of the procedure (see Section 2.1.5). Figure 2.1 illustrates the overall scenario of image-guided tissue ablation procedures.

The objective of tissue ablation procedures is the destruction of an area of pathological tissue inside the body. This area may be called targeted tissue volume [101], target volume [69, 196], target tumor [86, 32], target lesion [168, 206] or simply target [86]. In this work, the term target volume will be used. The ablation effect occurs in a small localized region called ablation zone\(^1\) [32]. This term is used to describe the tissue changes as depicted by medical imaging modalities or to refer to the expected zone of ablation. On the other hand, the volume of treated tissue resulting from pathologic findings is referred to as the coagulation. In this region all cells are necrotic as a result of the ablation. Although this term was used in the past mainly to describe the zone of cell death resulting from heat, it is also used to refer the zone of cell death created by newer technologies that do not rely on heat [201]. The various processes leading to coagulation are described in Section 2.1.3 and Section 2.1.4 in more detail. The distinction between coagulation and ablation

\(^1\)May also be called ablation area, active zone, or ablation volume.
2.1. Image-Guided Tissue Ablation Procedures

Figure 2.1: Abstract depiction of the ablation procedure scenario: The ablation device (blue) is positioned to facilitate treatment of the target volume (red). The means for tissue ablation are delivered by the generator (blue). An imaging device (yellow) is utilized to provide images during all phases of the procedure.

zone is necessary since pathologic findings are not necessarily identical to the zone of tissue changes visible in medical imaging [32]. In the scope of this work, the term ablation zone is used to refer to the probable zone of tissue necrosis for a given instrument.

2.1.2 Treatment Goals

The goal of an ablation procedure varies depending on the nature of the medical condition to be treated. Curative, and in many cases palliative (see Section 1.1) treatment of cancerous tumors typically requires the destruction of the whole tumor or even multiple tumors. Regions of undertreatment (also called underablation) must be prevented in order to destroy all pathological tissue. In addition to the complete target volume, a safety margin around it must be destroyed as well in order to eliminate tumor cells scattered in the immediate neighborhood of the tumor which are not visible to imaging [87],[58, p. 167]. For example, recommendations for margins around liver tumors vary between 5 mm [122] and 10 mm [142]. The size and shape of the coagulation may be influenced by parameters and physical properties of the generator and the ablation device as well as physiological properties of the surrounding tissue. If the target volume can not be covered by the coagulation resulting from one ablation, a series of ablations is required. However, since excessive ablation outside the target volume may impair function of the target organ [77] or could even injure adjacent structures [158], large regions of overtreatment (also called overablation) should be minimized as well. Hence, an ablation procedure is considered successful, if the whole target volume and the additional safety margin are covered by the coagulation and overtreatment is minimized. The relations between target volume, safety margin, ablation zone, undertreatment, and overtreatment are illustrated in Figure 2.2.
Debulking on the other hand, a palliative treatment form which aims at reducing tumor burden or controlling disease progression [32], does not require the destruction of the whole tumor. Uterine fibroids, for example, are often treated by reducing the mass of the fibroids by means of transcutaneous HIFU which induces large non-perfused areas in the tumors and therefore causes shrinkage of the fibroids [99].

### 2.1.3 Classification of Tissue Ablation Procedures

Tissue ablation procedures can be categorized based on the mechanism of action [32] that causes the tissue destruction as well as the means of delivering that mechanism to the target site. This categorization, which is detailed in the following, is summarized in Figure 2.3.

**The mechanism of action:** According to the classification introduced by Ahmed et al. [32], the mechanism of action can be either chemical or energy-based. This distinction is a recent update to the classification of Goldberg et al. [86], which categorized ablation procedures as either being chemical or thermal. However, due to the advent of more recent technologies (most notably irreversible electroporation (IRE), see Section 2.1.4.5), this scheme is not suited to encompass all modalities any more. Hence, the scheme by Ahmed et al., as well as recent literature use the term energy-based to describe all non-chemical modalities that utilize thermal or non-thermal mechanisms for tissue destruction [32, 42, 57, 201].

**Chemical ablation** procedures destroy the target volume by injection of liquid chemicals such as ethanol, acetic acid, and sodium hydroxide [206]. Therefore, these modalities rely on direct toxicity to produce necrosis [168]. The reaction between chemicals and tissue leads to cytoplasmic dehydration, denaturation of cellular proteins, and microvascular thrombosis which results in coagulation necrosis [206].

**Thermal ablation** procedures utilize the ablation device to destroy the tissue in the target volume by changing the temperature within that region. Hyperthermic ablation heats the target tissue and aims at acute coagulative necrosis. Generally, temperatures above 60°C lead to protein denaturation and melting of the plasma membrane which causes cell death nearly instantaneously. This temperature level,
which might vary depending on the cell type, is also described as being cytotoxic [113]. Although typically used to describe chemical or biological agents that kill cells, this term has been adopted to describe the quality or quantity of an ablative effect, that suffices to induce cell death [84, 87, 113, 151].

Hypothermic ablation cools the target tissue which leads to the breakdown of cellular metabolism, formation of ice crystals and osmotic shock. The cytotoxic range of hypothermic ablation is below $-40^\circ C$ [113]. In both, hyperthermic and hypothermic ablation, tissue destruction can also be caused by temperatures slightly outside the cytotoxic range. In this case, longer treatment times are required [113] since cell destruction depends on both the temperature and the time of exposure to that temperature [84]. As shown by Haemmerich et al. [89], it is important to incorporate the temperature history for an accurate determination of the lesion, especially outside the cytotoxic range. For hyperthermic ablation, the thermal dose is used to quantify the biological effect determined by the temperature and duration of the heating. Established modeling approaches include the Arrhenius formulation [95] and the CEM$_{43}$ (cumulative equivalent minutes of exposure at 43°C) model [169]. Similarly, for hypothermic ablation various ablative dose [42] models have been proposed. However, most publications mainly identified temperature thresholds [59, 144, 181]. A conclusive quantitative dose for hypothermic ablation depending on both, temperature and time has not been proposed, yet [42].

All thermal ablation procedures are negatively influenced by effects resulting from blood flow in the human body. The so-called heat sink effect describes the influence of nearby macroscopic vessels (larger than 1 mm in diameter) on the coagulation due to removal of heat [32, 82]. Similarly, warming by nearby vessels might lead to the cold sink effect during hypothermic ablation [86]. In both cases, the coagulation will be reduced in size and its boundary may be deformed. Additionally, capillary level microperfusion decreases the size of the coagulation and leads to an increase of the

Figure 2.3: Classification of ablation procedures with respect to the mechanism of action and the means of delivering the ablative effect.
necessary ablation duration \[127\]. Further negative effects resulting from the thermal nature include unwanted thermal injury of vascular structures or neighboring organs \[157, 138, 77, 110\].

**Non-thermal ablation** procedures use other principles than heating or cooling for localized tissue destruction. As a result, these methods do not suffer from the described adverse thermal effects \[120\]. Although the term non-thermal ablation is used in other fields as well, for example to describe certain methods for the treatment of varicose veins \[136\], it is only used to characterize one modality suited for the treatment of tumors: \textit{irreversible electroporation (IRE)} (see Section 2.1.4.5). Although, as pointed out by Ahmed et al. \[32\], classical chemical ablation procedures are also non-thermal, the term is not used in this context since the distinction into thermal and non-thermal has not been necessary for chemical ablation in the past. The emergence of \textit{thermochemical} methods \[61\] might, however, change this in the future \[32\].

**The delivery of the ablative mechanism:** The ablative effect can be delivered \textit{percutaneously, extracorporeally, endoscopically} or during open surgery \[113\]. This work focuses on the first two forms of delivery. \textit{Percutaneous} ablation procedures utilize needle shaped ablation devices to destroy tissue. As a result, these procedures are characterized by incisions of limited size and are therefore \textit{minimally invasive}. Depending on the modality, various terms are used for the device, for example, \textit{applicator, electrode, antenna, probe or needle}. For the remainder of this work, the more general term \textit{instrument} is used to refer to straight, needle shaped ablation devices. Typically, these instruments consist of a \textit{shaft} \[46, 196, 113, 151\] which is placed percutaneously into the patient body and a \textit{handle} \[46\], which is held by the physician (see Figure 2.4).

\textit{Extracorporeal} (also referred to as transcutaneous) ablation procedures utilize ablation devices outside the body to destroy tissue inside the body. No incision is necessary. Potentially, the skin as well as tissue layers between the skin and the target tissue remain unaffected. Therefore, extracorporeal procedures are also described as being completely \textit{non-invasive} \[99\]. The usage of this term is controversial as it mainly relates to the fact, that no physical object, e.g. instrument, penetrates the tissue of the patient. However, the energy, that leads to the ablative effect in the

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Figure 2.4: Illustration of percutaneous ablation setting.
target region, has to pass through the body. The fundamental idea of such procedures is, that only moderate quantities of the used energy transit through non-target tissue, which, ideally, leads to no noticeable effect in this tissue. Only the convergence of energy in the target region causes the therapeutic effect. Despite the controversy related to this term, it will be used synonymously to ‘extracorporeal’ in this work since both terms are commonly used. The only established non-invasive ablation modality, high-intensity focused ultrasound (HIFU), is described in Section 2.1.4.6. Figure 2.3 summarizes the presented categorization as well as the association of the various modalities, which are described in more detail in Section 2.1.4.

2.1.4 Energy-Based Tissue Ablation Modalities

The methods developed in this work focus on the support of percutaneous and extracorporeal energy-based ablation modalities. In the following, these modalities are described in more detail. The underlying principles as well as advantages and restrictions are explained.

2.1.4.1 Radiofrequency Ablation

RFA is the most popular and widely applied local tissue ablation modality [82, 133]. This percutaneous procedure realizes thermal ablation by applying a high-frequency (typically in the range of 375-500 kHz [86]) electric current in the target region. The alternating current flowing through the tissue causes ionic agitation and the production of frictional heat resulting from the resistance of the tissue. This leads to coagulative necrosis as a result of irreversible protein denaturation of the cells [113]. The electric field is delivered using instruments commonly called applicators. The electrodes responsible for creating the electric circuit are located close to the instrument tip, which is placed in the target tissue. Tissue closer to the electrode experiences a higher current and a greater rise in temperature then tissue farther away, which is mainly heated by thermal conduction [113]. Various electrode configurations have been developed over the last decades, including monopolar devices, which require one additional dispersive electrode placed on the patient skin. In addition to monopolar devices in the common needle-shaped form, several models using multi-tined expandable electrodes have been developed (see Figure 2.5 bottom). By dispersing the current over a larger volume, this design allows to increase the ablation zone size [113]. Bipolar systems implement a closed electric circuit with two electrodes, which might be placed on one or more instruments, allowing for multiple configurations depending on the number of available electrodes. The length, number, power, activation duration of and distance between the electrodes influences the size and shape of the ablation zone [205]. Depending on the used configuration, ablation zones with a size of 2-5 cm can be achieved [151] in typical treatment durations of 12-25 min [149, 147]. However, the size and shape is strongly influenced by the heat sink effect. This may result from the relatively long treatment times, allowing thermal energy to dissipate via blood flow [44, 151]. Several effects resulting from
the application of an electrical current, such as charring and tissue boiling, occur if the temperature is close to 100 °C. This leads to increased impedance of the tissue which limits further deposition of electricity into tissue [204, 196]. Various generator protocols and instrument designs have been developed to minimize this effect. For example, internally cooled instruments are used to limit the maximum temperature [204].

RFA is a broadly used modality. Besides hepatic malignancies, tumors in the the kidneys, breast, bone, and lungs have been treated clinically or experimentally [57]. RFA is indicated if no more than three lesions are present and the lesions are smaller than 3 cm in size [151]. Recent studies however hint at the possibility of treating larger tumors of up to 5 cm with a single instrument [57].

2.1.4.2 Microwave Ablation

*Microwave ablation (MWA)* is a percutaneous modality that uses electromagnetic waves in the microwave spectrum (300 MHz - 300 GHz, most commonly used frequencies for MWA are 915 MHz and 2.45 GHz [46]) to implement thermal ablation. Within the electromagnetic field polar molecules, such as water, continuously realign to the continuously changing waves. This oscillating movement generates the heating effect. All tissue within the electromagnetic field is heated simultaneously, which reduces treatment times in comparison to RFA [44]. Conduction of heat beyond the periphery of the field is a secondary effect [183]. Since microwave energy heats tissue so effectively [113, 206], the heat sink effect is less pronounced as in the case of RFA [204, 163].

The instrument delivering the electromagnetic field is called *antenna* and the radiating section close to its tip *radiator* [46] (see Figure 2.5 top). The design of antenna and radiator determines the shape of the electromagnetic field and the resulting ablation zone. Most designs for tumor ablation aim at a spherical field to match the probable tumor shape [46].

MWA is most often used to treat cancer in the liver, lung, kidney, prostate, and

**Figure 2.5:** Examples for percutaneous ablation devices: MWA antenna by Medtronic (Covidien Evident, top) and RFA applicator with deployable tines by Boston Scientific (LeVeen, bottom).
bone. Other target organs include the breast, brain, thyroid, and spleen [46]. Overall, the indications are similar to those for RFA [151]. However, MWA attains a more predictable ablation zone than RFA and slightly larger tumors can be treated (up to a size of 8 cm) [151]. The average treatment time per lesion is with approximately 4 min [151] much shorter than that of RFA. Furthermore, the presence of surgical clips or pacemakers is not a contraindication since a closed electrical circuit is not required [151]. A negative effect resulting from the high thermal efficiency is the possibility of injuring adjacent critical tissue [46, 151].

2.1.4.3 Laser Ablation

Percutaneous laser ablation, or also called laser-induced thermotherapy or laser interstitial thermal therapy (LITT), utilizes laserlight to induce heat in the target tissue. Typically, light generated by neodymium:yttrium aluminum garnet (Nd:YAG) lasers (wavelength of 1064 nm) [141, 195, 180, 113] or by diode lasers (wavelength of 800 - 980 nm) [141, 180, 64] inside the generator is delivered to the treatment site using fibers. While the fibers are flexible, a straight trocar is typically inserted into the tissue first and the fibre is placed inside it [206]. A scattering and diffusing tip at the end of the fibre is used to emit the light energy into the tissue where it is scattered and absorbed and eventually converted into heat [113]. The design of the tip plays a crucial role in defining the size [113] and shape of the coagulation (for example cylindric or ovoid [141]) [64]. Typical single ablation treatment times are in the range of 3 - 6 min [141, 64]. The maximum coagulation size is determined by the short travel distance of laser light inside tissue (12 - 15 mm) as a result of scattering, reflection, and absorption. Larger coagulation sizes result from further scattering of the light and conduction of the heat [206]. However, charring and dehydration might decrease the amount of scattering transmission of light energy [113]. Hence, the extent and completeness of the necrosis depends on a well chosen power setting [64]. Typically, ablation zones with diameters up to 3 cm can be achieved with a single fibre. Multiple fibers facilitate much larger ablation zones [206]. Laser ablation has been used to treat lesions in the liver [195, 64], lung [196], prostate [141], brain [137], and bone [80].

2.1.4.4 Cryoablation

Cryoablation, also referred to as cryosurgery, is the only hypothermic modality. It utilizes a percutaneously placed instrument called cryoprobe to lower the temperature in the target region in order to achieve cell death. A high-pressure gas (for example argon) is circulated through the cryoprobe. Low pressure within the tip of the cryoprobe results in rapid expansion of the gas, which creates a very low temperature (Joule-Thompson process) [113]. Due to convection and conduction, this low temperature is transferred to the surrounding tissue [206] which leads to the hypothermic effects described in section 2.1.3. Advantages of cryoablation are the relatively low amount of procedural pain [70, 133] and the faster and more complete
healing process after the procedure compared to hyperthermic ablation procedures [113]. Major drawbacks of cryoablation are relatively small [113] and variable [206] ablation zones, the cold sink effect [206], and high risk of hemorrhage and injury to adjacent organs [206] as well as the cryoshock [147, 113, 206], a systemic syndrome that is characterized by fever, tachycardia\(^2\), and tachypnea\(^3\) [206]. Applications of cryoablation include the treatment of cancers of the prostate, kidney, liver, breast, lung, bone, retina, and skin [42, 57].

### 2.1.4.5 Irreversible Electroporation

Irreversible electroporation (IRE) is a relatively new percutaneous non-thermal ablation modality. The needle-shaped probe is utilized to deliver ultra-short (20-100µs), high-voltage (1500-3000 V), high-current (25-45 A) [63] electrical pulses to the target tissue to destroy the cell membrane directly. The electric pulses create innumerable permanent nanopores in the cell membrane which initiate programmed cell death, the so-called apoptosis [120]. In addition to avoidance of the heat sink effect due to the non-thermal nature, this modality features several other advantages. The ablation times are very short: a single ablation takes less than one minute. As stated by Lee et al. [120] this may correlate with reduced anesthesia time, complications and post-ablation pain. In contrast to other ablation modalities, IRE spares vital structures such as vasculature within the ablated area [206]. Results of studies in swine [53] and humans [112] therefore suggest that IRE may be used to treat perivascular tumors (tumors close to vessels) that cannot be treated safely or effectively by RFA or MWA. Finally, the coagulation zones created by IRE feature sharp contours in contrast to necrosis resulting from thermal ablation. However, multiple IRE applicators need to be placed in a parallel manner to create moderate ablation sizes (two applicators are required to create a 2.5 cm ablation zone [113]) which increases the complexity, difficulty and the invasiveness of the procedure [100]. Furthermore, IRE must be performed under general anesthesia due to muscle stimulation and cardiac arrhythmias resulting from the strong, pulsed electric fields [63].

IRE has been used to treat tumors in the prostate, liver, kidney, lung, and pancreas in clinical studies [63, 100]. Of these sites, the liver seems to be the most promising target for this modality [100] while many studies indicate that IRE is not suited to treat tumors in the lung [63, 159], most probably due to a high sensitivity of the energy deposition of current IRE probes to air exposure [159]. Although it represents an interesting modality with unique properties, IRE is not an established treatment option yet and a better understanding of the method is required [63, 100].

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\(^2\)Tachycardia is a heart rate exceeding the normal resting rate.  
\(^3\)Tachypnea is an abnormally high breathing frequency.
2.1.4.6 High-Intensity Focused Ultrasound

Completely non-invasive tissue ablation can be achieved with extracorporeal high-intensity focused ultrasound (HIFU). Instead of the percutaneous utilization of needle shaped instruments the HIFU-transducer is placed outside the target region. No incision is needed as the transducer is placed outside the body (integrated in the patient table or held by a fixture or robotic arm, see Figure 2.6) or in a natural orifice. An example for placement of the transducer in a natural orifice is the transrectal treatment of prostate cancer. Even in this case, the treatment is still considered non-invasive [54] because the operation of the device does not require any incision.

The transducer creates beams of ultrasound energy which are focused onto the target. This target is also called focal spot. As individual beams travel through the intervening tissue, there is no effect until the beams converge in the focal spot which leads to profound non-thermal and thermal effects [99, 76]. HIFU uses moderate pressure and long pulse times which leads to strong thermal effects [99] resulting from the absorption of the acoustic energy [71]. A more thorough description of the physical principles of HIFU can be found in the review article by Schlesinger et al. [170].

The affected region, which is commonly called sonication (instead of using more general terms like ablation zone), typically has an ellipsoidal shape [99]. The term sonication is also used to refer to the process of carrying out the ablation and will also be used accordingly in this work. The sonications are rather small compared to the ablation zones of the previously described methods [71, 99, 111]. Typical single sonication sizes vary between $40 \text{ mm}^3$ and $300 \text{ mm}^3$, which corresponds to ellipsoids of dimensions between $8 \text{ mm} \times 3 \text{ mm} \times 3 \text{ mm}$ and $16 \text{ mm} \times 6 \text{ mm} \times 6 \text{ mm}$ [99, 167]. To treat larger target volumes, multiple sonications with varying focal spots are combined [99]. Typically, this so-called point-by-point method results in long treatment times due to the necessity of a cooling phase after each sonication [111, 197]. Treatment times in the range of hours are normal even for small and mid-sized

Figure 2.6: Illustration of extracorporeal HIFU: the transducer is positioned outside the body and focuses beams of acoustic energy at the focal spot inside the target volume.
tumors. Even a single sonication is a process of several minutes and treatments may need up to hundred sonications [170].

Modern HIFU-systems use multi-element phased array ultrasound transducers which consist of many small piezoelectric ultrasound elements. Each element can be parametrized individually with regard to the phase of the acoustic waves in order to steer the location of the focal spot in addition to mechanical positioning of the whole transducer [170]. The ability of electronic steering facilitates volumetric ablation, an alternative way to treat larger target volumes [71, 167]. Volumetric ablation methods steer the focal point along complex trajectories such as concentric circles [115, 71]. One goal of these strategies is the reduction of the necessary treatment time by minimizing or eliminating the cooling phases which are typically required for neighboring focal spots.

HIFU has evolved into an established modality with a broad range of indications. Approved oncological clinical applications of HIFU include treatment of bone metastases, tumors of the liver, prostate, kidney, and pancreas. Other approved indications are diseases in the fields of neurology (essential tremor, Parkinson’s disease, depression, obsessive-compulsive disorder (OCD), neuropathic pain) and women’s health (uterine fibroids, breast cancer, breast fibroadenoma, uterine adenomyosis) [74, p.4]. Until today, over 116,000 patients have been treated4. Prostate cancer (47%), uterine fibroids (31%) and liver cancer (14%) have been treated most often [74, p.6].

2.1.5 Stages of the Procedure and the Role of Image-Guidance

Image data of the patient anatomy play a crucial role in the preparation, execution and evaluation of any tissue ablation modality. As the first sentence of the book by Mahnken et al. [130] states, "every intervention starts with the visualization of the target organ or lesion and the path to the chosen target". Tissue ablation procedures can be broken down into several stages which contain various image-guided tasks. Goldberg et al. [86] as well as Jolesz [101] proposed breakdowns of the most important stages for needle-based ablation procedures. The system of Goldberg et al. , which has been recently updated by Ahmed et al. [32], also applies to HIFU [99]. Furthermore, Rueff and Raman [167] described the main phases of image-guided HIFU in a similar way. Hence, the following common main stages and subtasks can be defined for all tissue ablation procedures:

**Planning** is the process of determining a suited procedure based on preprocedural patient images. It encompasses the following subtasks:

- **Localization** [101]: identification of the exact spatial extent of the diseased tissue to be ablated. Furthermore, adjacent anatomical structures are identified, since they might be at risk of being injured during the ablative procedure [86, 32, 101, 99, 167].

4Case numbers as reported by manufacturers; real numbers might be higher [74, p.6]
• **Ablation planning**: determination of a suited placement of one or multiple instruments [101] or sonications [99] and all necessary device parameters [99].

**Procedure** refers to the process of implementing the planned ablations. **Intraprocedural** images are used to carry out the following subtasks:

- **Targeting** [86, 32]: the device is "aimed" at the planned position. For needle-based modalities image-guidance is used to enable the manual or guided placement of the instruments according to the treatment plan [86].

- **Monitoring** [86, 32]: the observation of ablation effects during the procedure. Changes in imaging are used to assess the completeness of target ablation as well as unwanted thermal injury.

- **Intraprocedural modification** [32]: the procedure is adapted to the current situation based on the information retrieved during monitoring. This may include repositioning of a device or changing of device parameters. This subtask is also known as **controlling** [86].

**Assessment** (of treatment response) [86, 32] refers to the evaluation of the treatment success after the actual procedure. For that purpose the position and size of the ablation zone as well as the coverage of the target volume by it are analyzed based on **postprocedural** images.

These stages and subtasks largely follow the proposal of Goldberg et al. [86] and Ahmed et al. [32], which, however, equalize planning with the identification and measurement of the target and anatomical structures only. However, as pointed out by Jolesz [101], structure identification and measurement is only the first part of planning, which Jolesz calls **localization**. The second planning step deals with the determination of a strategy for the actual ablations, which wasn’t considered by Goldberg et al. at all. Jolesz [101] calls this second step **targeting**, which however conflicts with Goldberg’s usage of that term. Hence, for the remainder of this work this step will be referred to as **ablation planning**.

To carry out the tasks in the described stages, imaging is utilized in three distinctive ways:

- The visualization of the anatomy and physiologic aspects of the target volume and the surrounding organs. This can be carried out using anatomic images resulting from ultrasound (US), computed tomography (CT) and magnetic resonance imaging (MRI) or physiologic images as produced by positron emission tomography (PET) or single-photon emission computed tomography (SPECT) [179, 32].

- The depiction of the instrument and its position in relation to the anatomy and the target volume. Continuous imaging can be achieved either using real-time or close to real-time imaging methods such as X-ray fluoroscopy, CT fluoroscopy, US, or intraoperative MRI [101]. Alternatively, conventional
volumetric imaging modalities might be used to create small 3D image volumes of the region of interest after each modification of the instrument position (sequential scans, [43, p. 17]).

- The verification of the ablative effect during and after the procedure. This can be either achieved considering ablation zone formation or tissue temperature distribution [211]. To some extent, CT, morphologic MRI and US can be used to assess ablation zone formation as the ablated tissue is slightly distinguishable from intact tissue in the images [101, 211]. For hyperthermic ablation modalities, thermometry imaging may be used to assess the tissue temperature distribution during the ablation. MRI thermometry exploits the temperature-sensitivity of various parameters such as relaxation times, proton density, proton resonance frequency, diffusion, and chemical shift [101, 153]. Similarly, many methods for US thermometry have been developed. However, they are only usable in a very restricted temperature range (41 °C - 45 °C) [211].

2.2 Planning of Percutaneous Tissue Ablation Procedures

In this section, the planning of percutaneous ablation procedures is discussed with a focus on ablation planning, which this work aims to improve. After summaries of possible complications and common planning considerations, the typical clinical workflow is described.

2.2.1 Complications

Complications of percutaneous ablation procedures can either result from the placement of the instrument or from the therapeutic effect [158, 138]. Instrument placement related complications include bleeding resulting from direct trauma caused by the instrument to vascular structures, pneumothorax, hemothorax, infections, and tumor seeding in the needle tract [158, 138, 77, 110]. Complications resulting from the therapeutic effect encompass unwanted ablation of non-target organs and vascular structures [158, 138, 77, 110]. Hyperthermic ablation of vasculature may lead to thrombosis. Smaller vessels tend to be more prone to thrombosis from thermal injury due to the reduced heat sink effect [110].

2.2.2 Planning Considerations

During planning of percutaneous ablation procedures, a suited instrument type, the access path of the instrument as well as device parameters such as applied power or ablation duration have to be determined such that a successful outcome (see Section
2.2. Planning of Percutaneous Tissue Ablation Procedures

2.1.2) can be achieved and previously described complications can be prevented. In the case of larger tumors, multiple access paths need to be defined.

**Instrument Model Selection** For each of the described modalities a range of different instruments from various vendors is available. They differ with respect to shaft length and the size of the achievable necrosis. Hence, the decision for a certain instrument type depends on the size of the target volume as well as its depth within the body, which have been determined in the localization phase.

**Access Path Selection** Accurate planning including the selection of a safe access path is essential for a successful intervention [202, 138]. The access path can be characterized by a **target point** and a **skin entry** point. The target point defines the position of the center of the expected ablation zone with respect to the instrument. It is located close to the tip of the instrument (see Figure 2.7). The distance of the target point from the instrument tip depends on the individual device design and must be learned from vendor specifications or from experience. The skin entry point is the point at which the needle penetrates the skin. However, any point on the shaft is sufficient to define the direction of the path (the term **shaft point** will be used for this more general case, see Figure 2.7).

To facilitate a positive outcome of the procedure, various aspects have to be considered during planning. In addition to achieving the treatment goals (see Section 2.1.1), the chosen strategy has to be as safe as possible. Therefore, a path should avoid penetration of certain anatomical structures along its trajectory [129, p. 56]. In this work, anatomical structures that must be spared are called **risk structures**.

![Figure 2.7: Complexity of the planning scenario illustrated for some of the considered criteria: The skin entry point (or shaft point) has to be defined such that distances to risk structures ($d_r$) are large enough to allow for a safe intervention. The distance between the skin entry point and the target point ($d_t$) should be as small as possible. However, the length of the passage through healthy tissue (transhepatic route, $d_h$) should not be too small. The instrument shaft should penetrate the liver capsule as close as possible to a perpendicular entry (therefore, $\alpha$ should be as small as possible).](image-url)
Incorporation of such risk structures during procedure preparation is, depending on the anatomical region, more or less complicated. For example, in the case of the liver this means that the lung, the costodiaphragmatic recess\(^5\) as well as several kinds of vessels (hepatic vessels, epigastric vessels\(^6\)) must be spared. Osseous structures such as the ribs as well as cartilage cannot be penetrated on principle. Furthermore, nerves beneath the ribs can cause considerable pain and neurological deficits if injured. Risk structures behind the target have to be taken into account as well since they might be damaged if the needle is advanced too far [43, p. 18] or due to the deployment of multi-tined expandable electrodes (see Figure 2.5 bottom). For hepatic interventions, the path should include enough healthy liver tissue (transhepatic route) to allow for cauterization of the path, for fixation of the needle and for a tamponade in the case of bleeding [129, p. 59], [157, 202, 110]. On the other hand, short paths are often preferred since long trajectories increase the risk of imprecision [38]. Furthermore, the maximal path length is restricted by the length of available instruments. Low liver capsule penetration angles should be avoided in order to prevent sliding on the liver surface and the resulting rupture [38]. Angulation of the path to the transverse plane\(^7\) should be as low as possible. Implementation of paths that are not in-plane (in the transverse image plane), so-called double oblique paths, can be more complicated depending on the used peri-interventional imaging device [43, p. 16]. Position and posture of the radiologist have to be considered in order to avoid impractical strategies. The same holds for the posture of the patient (for example, patients are typically in supine position during liver RFA [58, p. 171]).

**Multiple Ablations** In the case of large target volumes multiple ablations, either using one or multiple instruments, may be necessary. For example, some cryo systems allow to use up to 25 probes at the same time [168]. However, to reduce the overall risk of complications as well as the procedure duration, the number of ablations should be as small as possible [77]. The complexity of the planning process rises considerably if the target volume cannot be destroyed with a single ablation. If the target volume is slightly larger than a single necrosis, it may be possible to achieve complete destruction by using the so-called pull-back technique: multiple ablations are applied along the instrument trajectory [58, p. 222]. However, depending on the shape of the target volume and the direction of the instrument shaft, this approach might not always be sufficient and multiple overlapping ablations using multiple access paths have to be planned. The knowledge how to distribute the ablations to

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\(^5\)A potential space in the pleural cavity. The costodiaphragmatic recess as well as the lung are surrounded by the the pleura, which forms a sac. During inspiration, the lung partially expands into the costodiaphragmatic recess.

\(^6\)Vessels in the abdominal wall. For Liver interventions, mainly the superior epigastric vessels are of interest [190].

\(^7\)The transverse plane is an imaginary plane, which is oriented to the patient anatomy such, that it would be parallel to the ground if the patient is standing. Therefore, it divides the body into cranial and caudal portions. Alternatively, the term axial plane can be used. Images created by CT devices are typically in transverse planes.
2.2. Planning of Percutaneous Tissue Ablation Procedures

cover the whole target volume can be formalized using geometric patterns. Dodd et al. [66] propose instrument placement arrangements for 6 and 14 overlapping spheres placed on a regular grid. Chen et al. [56] and Khajanchee et al. [106] extend this model for larger target volumes. They use the centers of the faces of regular prisms [56] and regular polyhedra [56, 106] as ablation positions and introduce formulas that allow to calculate the number of needed ablations [56]. However, these models are based on the assumption of spherical tumor volumes and spherical ablation volumes.

2.2.3 Clinical Planning Workflow

In the clinical routine, the planning typically starts with the assessment of the target volume (localization phase). Its major diameter is measured using measurement tools which are available in any picture archiving and communication system (PACS). The required necrosis size limits the choice of suited instruments. As a consequence, the length of practical access paths is limited by the shaft lengths of the remaining suited instruments.

The access path is typically defined by means of interactive selection of target point and skin entry point (or another shaft point) in the axial 2D image slices. The resulting path may be visualized as a line. This can be achieved using simple tools such as measurement tools of a PACS. That way, the path length can also be compared with the length of the suited instrument types and the path can be adapted if necessary. If a path can be found that is in-plane, i.e. the target point and the skin entry point are in the same slice, this simple workflow may be adequate. However, often a path has to be chosen that is double oblique, i.e. the target point and the skin entry point are in different slices. Such angulated paths are necessary if no practicable path can be found in-plane without penetrating risk structures. For example, target points in the upper parts of the liver (sometimes referred to as liver dome) often necessitate strongly angulated paths. The definition of such paths typically requires the careful inspection of the path either by slicing through the 2D images or in multi planar reformation (MPR) images and checking for possible penetrations of risk structures by the previously defined line. If a collision is detected, a new skin entry point must be defined and the resulting path must be controlled again. This process has to be repeated until a satisfactory access path has been found.

Recently, several vendors of medical equipment introduced dedicated software for the support of percutaneous ablation procedures. Systems such as the Emprint™ Procedure Planning Application (Covidien/Medtronic, Boulder, CO, USA), XPerGuide (Philips Healthcare, Eindhoven, The Netherlands), ig4™ IR (Veran Medical Technologies, Inc., St. Louis, MO, USA), Ablation Confirmation™ Software (NeuWave Medical, Madison, WI, USA) and the MAXIO™ Planning Software (Perfint Healthcare Pvt. Ltd., Chennai, India) share common planning features such as:

- Import of patient data
- Identification of the target lesion
- Manual planning of virtual instruments in 2D and / or 3D
• Visualization of expected ablation zone (ellipsoids based on instrument vendor specification or manually adjusted)

These features correspond more or less to the state of research of 2003 [191]. Although no advanced planning tools like simulation of the ablation zone or automatic or semi-automatic selection of access paths under consideration of the patient anatomy are available in commercial solutions, the vendors have realized the need for advanced planning. However, no information is available on the degree of usage of such tools in the clinical practice.

2.3 Planning of High-Intensity Focused Ultrasound

Treatment planning is a critical aspect of HIFU that is necessary to avoid unnecessary, thermal damage in normal tissue [93] and to determine parameters that facilitate a successful procedure (see Section 2.1.2). In the following, aspects of the planning of HIFU procedures are described after a summary of the most common complications.

2.3.1 Complications

A possible complication encountered during HIFU is the unwanted heating of the skin (skin burns) [185, 170]. Furthermore, gas filled regions such as the bowel and bony structures such as the ribs cannot be passed by HIFU and therefore cause acoustic shadows. As a result, these structures might be heated due to absorption, or other structures in proximity might be heated as sound waves are reflected back [185, 67, 37, 170].

2.3.2 Planning Considerations

During HIFU ablation planning two major tasks have to be solved. First, the position of the transducer with respect to the patient has to be determined. Second, individual sonications have to be defined in order to reach the treatment goals.

**Transducer placement:** Depending on the used HIFU system, the transducer can be positioned and oriented with regard to various degrees of freedom. During planning, a positioning has to be determined which facilitates the propagation of acoustic energy to the target and which prevents the previously described complications. Hence, an adequate *acoustic window* has to be guaranteed [185, 67]. Furthermore, efficient focusing, and thus ablation, is only possible in a certain distance from the transducer. Several transducer designs have a fixed aperture and fixed focal length [99]. Hence, the transducer position and rotation also determines the focal spot. HIFU-systems with multi-element phased array ultrasound transducers allow to steer the position of the focal spot. However, the overall configuration of the elements follows a concave shape in most cases in order to facilitate a natural
focus [170]. Hence, also these systems have to be positioned such that the natural focus is close to the target region. Finally, the energy level between the transducer and the target (near-field), as well as behind the target (far-field) should be as low as possible in order to prevent complications such as skin burns. This can be achieved by placing the transducer such that the resulting ultrasound beams have a wide angle of aperture [185, 67].

Sonication planning: Typically, sonication are much smaller than the target volume to be treated. Hence, coagulation zones of sufficient size need to be created by combining many sonication. This so-called point-by-point method requires the planning of many independent sonication such that the resulting combined coagulation covers the target volume completely. This process is further complicated by the fact, that the exact position, shape and size of the coagulation can not be easily predicted. The size and shape depend on the geometry and parameters of the transducer as well as the absorption of the tissues encountered along the beam path [47]. The apparent focal spot may be shifted as a result of phase aberrations caused by tissue inhomogeneities [93].

2.3.3 Clinical Planning Workflow

The non-invasive and highly technical nature of HIFU requires detailed, software-assisted treatment planning and execution, since there is no way to implement the therapy manually. Configuration of the transducer and execution of the therapeutic ultrasound delivery is completely software controlled and requires sonication parameters as input.

There are two clinical magnetic resonance (MR)-guided HIFU systems on the market: the ExAblate (Insightec, Israel) and Sonalleve (Philips Healthcare, Finland) systems. Both systems utilize multi-element phased array transducers and feature computer-assisted planning tools, which allow to localize the target volume and to define beam paths and sonication positions. The placement of sonication can be restricted by so called no-pass zones in order to spare sensitive structures [170]. A description of the general workflow in the ExAblate planning software is given by Foley et al. [76]: the software facilitates manual segmentation of the target volume and automatically computes a treatment plan. The individual sonication can be manipulated by the user. Furthermore, the beam path for each sonication can be superimposed on the MR image data in order to control it with respect to focusing through air cavities or highly absorbing and reflecting bony structures. If necessary, the beam path can be adapted manually. This process can take up a significant fraction of total treatment time[170]. The planned sonication are automatically implemented by a combination of mechanical transducer placement for gross positioning of the focal spot, and electronic steering for fine control [170].

Among single element transducers, the products of HAIFU Medical (Chongqing, China) are broadly used [74]. These US-guided systems implement sonication by positioning and rotating the transducer with respect to six degrees of freedom. The
planning software provides a targeting module for the delineation of the target volume and a planning module for sonication definition. For specific devices, such as the transrectal Sonablate (SonaCare Medical, USA) or the Ablatherm (EDAP TMS S.A., France) systems for prostate cancer treatment, the planning workflow might differ slightly.

2.4 A Short Introduction to Computer Graphics and Medical Visualization

Computer graphics allow to create images, both still and moving, of virtual objects with the help of computers. Two decades ago, this ”esoteric specialty involving expensive display hardware, substantial computer resources, and idiosyncratic software” [75, p.XI] started its transition to a mainstream technique that today penetrates science, education and entertainment. This evolution has been made possible mainly by the development of specialized hardware, the so-called graphics processing unit (GPU). These highly specialized, parallelized and programmable processors allow for real-time computation of very realistic images.

In the following, several concepts from the field of computer graphics and visualization important for medical visualization in general, and in particular for the work presented in this thesis, are described.

2.4.1 Three-Dimensional Rendering of Volumetric Medical Data

The prevalent techniques for interactive rendering of medical data sets are surface rendering and volume rendering. In surface rendering, the objects to be displayed are modeled by polygonal surfaces. In the context of medical image processing and visualization, which is typically concerned with volumetric image data (see Section 2.1.5), an additional polygonization step is required, typically [154, p.160]. It extracts isosurfaces from volume data and represents them as polygonal data structures which can be displayed using surface rendering.

Volume rendering on the other hand, allows to display volumetric data directly and therefore displays characteristics of the interior of an object. For that, the volumetric data is sampled along the viewing ray. The sampled data values of the volume are mapped to optical properties (i.e. color and opacity) by means of transfer functions [154, p.160]. Various methods for the aggregation of the samples along the ray allow for different effects. Direct volume rendering (DVR) accumulates the samples along the ray incorporating their opacity. This allows to show the various tissues as layers of different opacity and color and therefore facilitates a realistic depiction of the morphologic information. Maximum intensity projection (MIP) and minimum intensity projection (MinIP) on the other hand show the maximum and minimum value along the ray, respectively [209].
2.4.2 The GPU and the Graphics Pipeline

The original rendering pipeline of the GPU was static (fixed-function pipeline) and was mainly tailored towards surface rendering. However, it changed over the years and various new stages have been added (see [41, p.25-37]), some of which can be programmed using shading languages such as the OpenGL Shading Language (GLSL) [109]. Figure 2.8 shows the graphics pipeline for OpenGL [109], a common real-time graphics application programming interface (API) (the stages are similar in other APIs). The vertex shader allows to execute computations on the vertices that are passed to the GPU by the application. It has to implement all transformations that are applied to objects in the scene and maps the geometry into the coordinate system of the camera. In addition, it might be used for example to deform objects. In the primitive assembly steps, primitives such as lines or triangles are generated based on the processed vertices. Tesselation shaders subdivide the input geometry and therefore allow to generate surfaces of smoother appearance and with more detail. Geometry shaders, on the other hand, are used to create completely new primitives based on the input geometry. The rasterizer interpolates the vertices to create fragments. A fragment is a part of a primitive that is projected onto a pixel and therefore potentially contributes to the color of that pixel. During this interpolation process, vertex attributes such as depth, color, and texture coordinates are interpolated as well. Finally, the output color of a fragment can be further modified using fragment shaders, which are also called pixel shaders sometimes (for example in DirectX, a graphics API similar to OpenGL).

This programmability gave rise to various real-time methods for DVR such as texture mapping, raycasting, splatting, shell rendering, cell projection and shear-warp [209].

2.4.3 Visibility Determination

An important aspect of the rendering of three-dimensional scenes is the determination of the visibility of surfaces from a certain point, the so called visible-surface

Figure 2.8: The OpenGL graphics pipeline with programmable stages shown in blue and fixed-function stages shown in gray. Illustration adapted from [41, p.40].
determination or hidden-surface removal problem [75, p.649]. Two popular and broadly used approaches for solving that problem are raycasting and the z-buffer algorithm. Raycasting explicitly computes intersections of each viewing ray with the objects in the scene to determine which surfaces are visible on a certain pixel [75, p.701]. The z-buffer algorithm is embedded into the rasterization process carried out by the GPU. It uses an additional buffer, the z-buffer or depth buffer, that keeps track of the depth of the surface fragment at a certain pixel that is closest to the camera. During rendering this information is updated for each object being rendered and each pixel. Hence, a surface is visible at a certain pixel if its depth is not larger than the depth of the surfaces that have been projected onto that pixel before. Then, it overwrites the color in the output image and the depth in the z-Buffer [75, p.668]. At the end of the rendering process, the z-buffer contains for each pixel a measure for the depth of the closest visible surface fragment that is projected onto that pixel. Typically, this measure is not linear due to the involved perspective projection.

2.4.4 The Cube Map

Another central concept for many real-time rendering techniques is the so called cube map, which consists of 6 perspective projections. Each projection uses a perspective camera that is centered at the projection center and has a vertical and horizontal field of view of 90°. The cameras are oriented along the positive and negative directions of the three coordinate axes. This setup allows to capture all directions for the projection center. Hence, cube maps are omnidirectional. The cube map concept is illustrated in Figure 2.9. Cube maps were originally developed for environment mapping, a method to approximate reflections in real time rendering [88], and omnidirectional shadow mapping, a method for real time shadow generation [83]. In both cases, the cube map is computed once before the actual rendering and then used during rendering to display the wanted effect on the object surfaces. Creation of the cube map can be done with the same rendering system that would create the final image [88]. Hence, in real time rendering systems the cube map may also be generated at interactive frame rates. However, it can be reused for multiple renderings until the objects that are captured in the map, or the projection center, have changed. Only then a recomputation of the map is necessary. Once the cube map has been generated, it is accessed using a three-dimensional lookup vector \(v\) which represents the direction from the projection center for which a value should be fetched from the map. The result of evaluating a cube map for \(v\) approximates the result of casting a ray defined by the projection center and the direction of \(v\). During evaluation of the map for \(v\), the largest coordinate of \(v\) determines from which of the 6 faces of the cube map to sample. The remaining two coordinates are divided by the largest coordinate and then used to determine the sample position in the selected face [121, p. 171]. Typically, this texture lookup functionality is optimized and exposed by shading languages such that only the
vector $v$ has to be specified. For example, in GLSL, a value in the direct $v$ is fetched from `cubeMap` using the function `gvec4 texture(gsamplerCube cubeMap, vec3 v)`.

### 2.5 A Short Introduction to Decision Making and Numerical Optimization

The goal of numerical optimization is to identify a variable $x \in \mathbb{R}^n$ for which the value of a function $f(x)$ is *optimal*, thus minimal or maximal. Based on an initial guess of $x$, a sequence of consequently improved estimates is generated, hopefully terminating with an optimal solution \[146, \text{p.8}\]. The definition of the optimum is straightforward for a scalar function as the scalar values of $f(x)$ can be sorted using the relation *less than or equal* ($\leq$). Hence, the minimum or maximum define the optimum, and optimization refers to minimization or maximization of $f(x)$, respectively. Commonly, optimization is formulated as minimization: problems requiring maximization can be solved by minimizing $-f(x)$. In the case of a scalar function, an optimization problem can be formulated as follows:

\[
\begin{align*}
\text{subject to } g_j(x) &\geq 0, & j &= 1, 2, \ldots, m, \\
& & h_l(x) &= 0, & l &= 1, 2, \ldots, e, \\
& & f, g_j, h_l: \mathbb{R}^n &\rightarrow \mathbb{R},
\end{align*}
\]

\(2.1\)

where $m$ and $e$ are the number of equality and inequality constraints, respectively \[132][146, \text{p.2-3}\]. The set of all vectors $x \in \mathbb{R}^n$ that fulfill all equality and inequality
constraints is called the feasible design space, feasible set or feasible decision space \( \mathcal{X} \). Its elements may be called vector of design variables, vector of decision variables or point of the design space [132]. This formulation allows to describe a basic single-objective optimization (SOO) problem. However, in many real-world problems, decision makers need to reach an optimum under consideration of a number of decision criteria. Hence, \( k \) objective functions \( f_i(x) \) have to be considered. Usually, at least some of them are contradictory. These kinds of problems are called multi-criteria decision making problems [186]. Multi-objective optimization (MOO) (also called multi-criteria optimization) can be used to solve such problems. The definition of the optimum in MOO is not as straightforward as in the case of SOO, since there is no canonical order on \( \mathbb{R}^k \) [98, p.30]. Typically, a single optimal solution does not exist and multiple points that can be regarded as optimal with respect to a predetermined definition of an optimum have to be found [132]. The prevalent concept for the definition of optimal points in MOO problems is the so called Pareto optimality: A point, \( x^* \in \mathcal{X} \), is Pareto optimal if it is not dominated by any other point \( x \in \mathcal{X} \). Therefore, there does not exist another point, \( x \in \mathcal{X} \), such that \( f_i(x) \leq f_i(x^*) \), and \( f_i(x) < f_i(x^*) \) for at least one function [36, 132]. Hence, a point is Pareto optimal (also called Pareto efficient) if no objective characterizing that point can be improved without deteriorating some other objective. The set of Pareto optimal points in \( \mathcal{X} \) is called the Pareto set. The corresponding vectors of objective function values in the criterion space lie on the boundary of \( \mathcal{Z} \) and form the Pareto front (see Figure 2.10). Thus, the goal of MOO is to find the Pareto front.

Analogous to Equation 2.1, the MOO problem can be formulated as follows:

\[
f(x) := [f_1(x), f_2(x), ..., f_k(x)]^T \rightarrow \min
\]

where \( f(x) \) is the vector of the \( k \) objective functions \( f_i(x) \). The feasible criterion space \( \mathcal{Z} \), also called the feasible cost space or attainable set, is defined as the set \( \{f(x) \mid x \in \mathcal{X}\} \) [132]. Figure 2.10 illustrates the relation between \( f \) and the two spaces \( \mathcal{X} \) and \( \mathcal{Z} \).

Approaches for the determination of Pareto optimal points in MOO problems can be categorized based on the way the decision maker articulates his or her preferences [36, 132]:

- **A priori articulation of preference information:** These methods allow to specify the preferences of the decision maker in terms of goals or the relative importance of the different objectives. Hence, before the actual optimization is carried out, the different objectives are aggregated to one single figure of merit based on these preferences. In utility theory, a utility function \( U(f(x)) \) is used to map \( f(x) \) to a scalar based on the utility that a certain solution has to the decision maker. Other examples include weighted sum approaches, non-linear combination, fuzzy logic approaches, acceptability functions, the lexicographic approach, and goal programming.

- **No articulation of preference information:** These methods do not use any information about preferences of the decision maker. Examples are the
Nash arbitration scheme, Rao’s method and global criterion methods. These methods are often simplifications of methods with a priori articulation of preferences typically missing any parametrization [132].

- **Progressive articulation of preference information**: These methods use preference information which is refined simultaneously while the search for the optimum is performed. Typically, after each iteration of the optimization process, some solutions are presented to the decision maker who has to select the solution which best fits his preferences. This information is incorporated during the next iteration to generate new solutions [103]. Hence, these methods are also referred to as interactive methods. An overview of according approaches can be found in [103].

- **A posteriori articulation of preference information**: The decision maker selects from a list of possible solutions after the optimization has been finished. One approach is to use an a priori method like the weighted sum approach and run the optimization multiple times with varying weights (multiple run approach [36]) where each run generates one sample of the Pareto front.

Many of these approaches follow the idea of scalarization: the MOO is reduced to a SOO by aggregating the multiple objectives into a single scalar function. Hence, the optimization itself can be carried out using classical optimization schemes, which can be divided into derivative and non-derivative (or gradient-based and gradient-free) methods [36]. Derivative methods, like Newton’s method, use derivatives of $f(x)$ to define a search direction. Typically, these approaches are well suited to solve local optimization problems efficiently [78]. However, depending on the application, an
analytical computation of derivatives is not feasible or computationally expensive and an approximation based on finite differences might be required [146, p.220]. Non-derivative methods do not require derivatives of \( f(x) \). Hence, \( f(x) \) can be treated as a \textit{black box} method [36]. These approaches are well suited to narrow down the area of global optima. However, they are often inefficient for the exact determination of the local minimum depending on the chosen step sizes [78]. Examples are \textit{Powell’s method} [152], the \textit{Nelder-Mead Simplex}, \textit{simulated annealing}, \textit{genetic algorithms} [36] and \textit{brute-force} approaches. The latter simply tries to compute \( f(x) \) for as many \( x \) as possible and selects the optimum afterwards [108, p.181].

As both classes have individual advantages and disadvantages, they are often combined into \textit{hybrid} methods [36, 78].

### 2.6 Contributions to the State of the Art

The methods proposed in this thesis aim at the improvement of the described planning workflows. For that, techniques from various fields are utilized and combined. Especially computer graphics play an important role in each of the developed methods. Both, access paths and light rays can be modeled as lines. Hence, computer graphics algorithms, that originally aimed at the modeling of light and visibility, are exploited here for the determination of suited access paths. The cube map is utilized together with DVR to allow for GPU-accelerated analysis of the patient anatomy. Chapter 3 presents a visualization method that uses this approach to highlight infeasible access paths directly in the established 2D visualizations in an interactive manner.

The first optimization method presented in this thesis (Section 4.4) uses an a priori articulation of preference information based on utility theory to scalarize the results of a brute-force computation of the objectives using the cube map. This method computes clinically relevant access paths almost instantaneously. In the second method, this approach is used in a multiple run manner to yield starting points for a hybrid numerical method which uses the \textit{adapted hyperboxing} method by Teichert [184] to calculate a representative set of efficient solutions. These are interpolated based on interactive a posteriori articulation to yield one selected solution [118].

The resulting method is the first MOO approach that considers a plausible model of the ablation zone including heat sink effects in addition to other clinically relevant criteria and allows for interactive exploration of the space of solutions.

In Chapter 5, a method for the interactive approximation of the heating resulting from HIFU is presented. It utilizes image processing to derive compact sonication representations from numerical simulation results for a large number of exemplary sonications which are stored in reusable lookup tables. During interactive planning the heating for a planned sonication is interpolated from these lookup tables. The reconstruction of the heating pattern based on the interpolated information is carried out on the GPU using shaders.
3 Interactive Feasibility Visualization for Access Path Planning

Once you eliminate the impossible, whatever remains, no matter how improbable, must be the truth.

(Sir Arthur Conan Doyle)

Identification of a feasible access path for the instrument is a crucial step in the planning of an intervention. The established planning workflow is based on viewing 2D slices of a pre-interventional CT or MR scan and requires the careful inspection of each candidate path with respect to collision with risk structures and other criteria. In this chapter methods are presented which augment the definition of access paths in two- and three-dimensional visualizations. Projections of obstacles such as risk structures computed using the graphics processing unit (GPU) are used to highlight paths that would penetrate any of these structures.
3.1 Introduction

Percutaneous ablation procedures need to be planned carefully. The physician has to choose a linear access path, which does not penetrate any risk structures. Furthermore, only paths with a length that does not exceed the instrument length are feasible (see Section 2.2.2 for more details on planning considerations). Today’s standard planning workflow is still mainly manual and based on the inspection of two-dimensional image slices (see Section 2.2.3 for a detailed description). If a path can be found that is in-plane, i.e. the target point and the skin entry point are in the same transverse slice, this simple workflow is adequate. However, often a path has to be chosen that is double oblique, i.e. the target point and the skin entry point are on different slices. Such double angulated paths are necessary if no practicable path can be found in-plane without penetrating risk structures. For example, target points in the upper part of the liver (liver dome) often necessitate strongly anguluated paths. The definition of such paths typically requires the careful inspection of every candidate path by scrolling through the 2D images and checking for possible penetrations of risk structures by the previously defined line. Furthermore, the length of the path has to be taken into account. If any of these two constraints is violated, the skin entry or target point must be modified and the resulting path must be controlled again. This iterative process constitutes a demanding mental task, is time consuming and requires a considerable amount of experience. Therefore, the process of access path planning for percutaneous tissue ablation procedures can strongly benefit from computer assistance.

Many publications aiming at software support for access path planning focus on the depiction of the patient anatomy including relevant risk structures in three-dimensional visualizations. The resulting environment allows for positioning of virtual instruments and for visual inspection of the resulting paths with regard to violation of any of the mentioned criteria. However, only few works aim at supporting the task of access path definition directly in the 2D image slices, which radiologists are trained to gather information from [102].

In this chapter, novel methods are presented that enable the physician to identify feasible instrument placements in a fast and safe manner. The placement of an instrument is defined by manipulating two points, the target point and the shaft point of the instrument (see Figure 3.1). The aim of the developed methods is to highlight all points, that would define target or shaft points of infeasible paths, i.e. paths that would penetrate a risk structure or an already existing instrument, or that would be too long for the chosen instrument type. These points are highlighted directly in 2D image slices of anatomical images from CT or MR scans, or on the skin of the patient in three-dimensional renderings.

Contributions of this work include:

- Proposal of an efficient method for the visualization of infeasible shaft points in 2D viewers based on projections of risk structures utilizing the GPU.
3.2 Related Work

- Advancement of the previous approach to a comprehensive method for visualization of infeasible access paths in both, 2D and 3D viewers. It extends the first method to also augment the definition of the target point and to incorporate collisions of the instrument tip. Furthermore, instruments that have already been planned are considered as obstacles and an additional safety margin as well as the restriction of the path length are incorporated.

The chapter is structured as follows: Section 3.2 summarizes related work in the field of planning of tissue ablation procedures. Section 3.3 and Section 3.4 present the two methods for visualization of infeasible access paths. Finally, in Section 3.5 the proposed methods are discussed in the context of comparable work.

### 3.2 Related Work

Multiple software applications for the support of procedure planning for RFA [193, 134, 200, 105], MWA [208, 125], cryoablation [50] and laser ablation [124] have been described in the literature. An obvious and often used solution to augment the search for a suited access path is the utilization of 3D rendering techniques to illustrate the anatomical situation. Generally, the goal of the visualization is to allow for a better understanding of the individual patient anatomy and to support the physician during the determination of a suitable intervention strategy by incorporation of additional planning information. Such information may include image masks representing anatomical structures of interest. These masks are the result of manual or automatic image segmentation methods. In most publications, the anatomy is visualized in a 3D viewer either using surface rendering [50, 193, 125] or direct volume rendering (DVR) [134, 200, 208, 162, 105] (see Section 2.4.1). The interactive modification of the position and orientation of the virtual camera allows to investigate the anatomical situation and to spot suited access corridors. In addition to the 3D viewer, axial, coronal and sagittal viewers show the patient images and overlay the extracted anatomical structures. Virtual instrument models based on vendor specifications may be placed interactively in these visualizations. Estimations of the expected ablation

![Figure 3.1: Illustration of instrument placement definition using target and shaft points.](image)

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zones are superimposed onto the anatomical information to allow the physician to check if the type of the selected instruments as well as their placements are suited to destroy the target volume. To this end, ablation zones are either represented as ellipsoids based on vendor specifications [50, 134, 29, 162, 182, 125] or numerical simulation algorithms are utilized to display a prediction based on the modeling of the underlying biophysical processes [124, 200, 208, 105]. A few works propose approximative approaches to represent ablation zones including the heat sink effect in an interactive manner [193, 116, 161, 12].

The process of finding suited access paths can be further augmented based on the knowledge about the patient anatomy. McCreedy et al. [134] display the intersection of the defined path with any segmented critical structure as well as the skin entry point. While this allows to assess a path that has already been selected, it does not facilitate an overview of other alternatives. Such an overview may be realized by projection of feasible regions or path quality criteria onto the outer surface of an anatomical rendering. Villard et al. [192] proposed to visualize insertion zones suited for liver-RFA on the patient skin. All points on the surface, that would define a path that does not penetrate a critical structure, are highlighted. This approach is extended to include additional criteria such as penetration depth, liver capsule penetration angle and penetration of healthy liver tissue [39]. Each of these criteria is interpreted as a hard constraint, resulting in a binary segmentation of the surface. The authors describe the combination with qualitative measures in a concurrent publication [38]. Similar approaches have been developed in the field of neurosurgery planning. Vaillant et al. [188] visualize the risk of each path on the brain surface as well as on a overview map based on a parametrization of the surface. The computed risk evaluates which structures are penetrated. Many following works in the field [48, 143, 174] describe approaches that are very similar to the ones proposed by Baegert and Villard for RFA: They compute and visualize path cost based on the distance to critical structures and overlay it onto the brain surface. As an additional criterion penetration depth is considered by some works [143, 174]. The computation times are in the range of 15 minutes. To speed up the computation, Herghelegiu et al. [96] propose to restrict the computation to an ellipsoid region on the surface which is centered at a user-defined position.

A completely different approach to visualize path suitability for planning of interventions in both, liver and brain, is presented by Khlebnikov et al. [107]: they propose to model the tumor as a volumetric light source and visualize the resulting crepuscular rays in a three-dimensional volume rendering. The user sees the amount of light that is emitted and not blocked by any other anatomical structures. This visualization is inspired by the volumetric shadows often encountered in reality on foggy days. It is accompanied by a simplified 2D version: the precomputed crepuscular rays-volume is used to displace a surface mesh of the target volume. Each vertex of the mesh is translated along its normal vector until an unsafe voxel is encountered in the crepuscular rays-volume. The authors conducted an evaluation study, which showed that acceptance of this simplified 2D version is much higher than for the three-dimensional volume rendering. However, only few investigators focus on improvement of 2D
viewers, which radiologists are used to. Path-aligned MPRs, which are common for tracking systems, can also be used during planning to check a selected path [162]. Herghelegiu et al. [96] augment instrument path display in MPRs by highlighting the closest vessel.

### 3.3 Display of Infeasible Shaft Points in 2D Slice-Based Visualizations

The aim of the method proposed in this section is to augment the selection of shaft points in 2D viewers. It is assumed that the target point can be easily defined by the user and only the definition of the second point, the shaft point, requires visualization support. The method supports shaft point selection by highlighting all points, that are not feasible as shaft points since they would define a path that penetrates a risk structure. To this end, risk structures to be taken into account are given by one unified mask image (risk structure mask) which classifies all voxels as either belonging to a risk structure (value 1) or to the background (value 0). The visualization has to be updated interactively while scrolling through the stack of 2D image slices resulting from CT or MRI scans. To achieve that, the GPU-computed risk structure map is introduced. The overall algorithm consists of two stages:

1. **Risk structure map generation**: Computation of the risk structure map
2. **Visualization**: Projection of the risk structure map

Both stages utilize the GPU and are described in the following.

#### 3.3.1 The Risk Structure Map

A point \( p \) is an infeasible shaft point if the line segment between \( p \) and the target point \( t \) penetrates a risk structure (see Figure 3.2 (a)). Detection of intersections between lines and objects have been heavily studied in the field of computer graphics, especially in the context of visible surface detection (see Section 2.4.3). Raycasting could be used to solve the described intersection detection problem. However, a large number of computations are necessary if this approach is expanded onto a whole 2D slice or even multiple reformations (see Figure 3.2 (b)). Depending on the used hardware this can be too expensive. Hence, the utilization of an intermediate risk structure map is proposed. It allows to precompute and cache penetration information in order to enable a real-time 2D visualization. The risk structure map is based on the cube map concept (see Section 2.4.4). The rendering into the cube map is modified as follows: Each pixel needs to store if the path that passes through that pixel is infeasible, therefore if it penetrates any segmented risk structure. With respect to the cube map this correlates with the visibility of objects. Thus, pixels that show the surface of any risk structure correspond to infeasible paths. In the risk
Chapter 3. Interactive Feasibility Visualization for Access Path Planning

Figure 3.2: A point $p$ belongs to an infeasible path regarding the target point $t$, if the line segment between $p$ and $t$ penetrates a risk structure (left). If applied to all points on a slice (horizontal lines, right), on the skin or in 3D space, whole sectors of infeasible paths can be identified (red, right).

Structure map, the value 1 is assigned to these pixels. All other pixels have the value 0. Practically, DVR is used to render the risk structure mask into the risk structure map. Illumination is not computed and all structures are rendered in plain white. The map is updated subsequently to any target point modification. Figure 3.3 shows example risk structure maps for two different sets of risk structures.

Figure 3.3: Exemplary risk structure maps: (a) shows the risk map for ribcage and lungs while (b) also contains the projection of liver vessels.


3.3. Display of Infeasible Shaft Points in 2D Slice-Based Visualizations

3.3.2 Visualization

Every pixel of a given slice has to be highlighted if it belongs to an infeasible path. To achieve that, the previously computed risk structure map is projected onto a rectangle that represents the displayed image slice in 3D space. This rectangle is rendered over the 2D slice containing the patient image data. A shader is used to modify color and transparency of the rectangle based on the projection of the risk structure map onto the rectangle. If the result of a projection is 1 for a certain point, the point belongs to an infeasible path and is displayed in opaque red. For all other points the rectangle is completely transparent. Thus the original gray values of the underlying patient image data are displayed. To reduce the occlusion of the original image data, the visualization is restricted to points outside of the patient contour based on 3D texture containing the patient mask. The combination of this masking process with the projection is outlined in Algorithm 1.

To allow for a completely unoccluded display of the original image data, the risk structure projection is completely hidden as soon as the user moves the mouse pointer to a point inside the patient or if the mouse leaves the viewer.

Algorithm 1 Shader pseudo code for visualization of infeasible shaft points: each point \( p \) is classified with respect to the target point \( t \) based on the risk structure map \( \text{riskMap} \).

```
function HIGHLIGHTINVALIDPATHS(p, t, patientMask, riskMap)
    v ← p − t
    inside ← textureLookup(patientMask, p) \( \triangleright \) 1: inside patient, 0: otherwise
    collides ← cubeMapLookup(riskMap, v) \( \triangleright \) 1: collision, 0: otherwise
    if not inside and collides then
        return color(1,0,0,1)
    else
        return color(0,0,0,0)
```

3.3.3 Results

The described method has been implemented utilizing MeVisLab [164], a freely available development platform for medical image analysis, visualization and application prototyping. The built-in GigaVoxelRenderer [123] has been used to render the risk structures into the risk structure map. The projection of the map onto the 2D slices has been implemented using GLSL, which supports hardware-accelerated sampling of cube maps.

The proposed visualization method has been integrated into a software prototype for the planning of hepatic interventions such as biopsy and RFA based on CT image data. This prototype offers automatic segmentation of the bones, the lung and other air filled parts (stomach, bowel). Depending on the contrast phase, the kidneys might be segmented too. This preprocessing step is carried out in less than
30 seconds (Core 2 Duo E8500 @ 3,16GHz, 2GB Ram, NVIDIA GeForce 8800 GT, Windows XP 32bit). In the software prototype the user defines the target point manually by selecting an appropriate slice and by clicking into the image at the desired position. The target may also be defined automatically based on prior tumor segmentation. However, manual definition should be possible in any case since the tumor center cannot be considered to be the optimal target in all cases, e.g. for larger tumors. Based on the preprocessing results and the user defined target point, the risk structure map is computed in less than one second. It is only updated after initial definition or modification of the target point. The projection of the map onto the 2D slices is carried out in real-time. Hence, slicing through the 2D images is not slowed down compared to standard slicing.

The visualization can be applied to different views and reformations simultaneously since the target point and the risk structure map are independent of the view or reformation.

3.3.4 Evaluation

To assess the clinical value of the proposed visualization method a retrospective evaluation study is carried out. The following hypotheses are evaluated:

- **H1**: The proposed method supports the definition of feasible access paths.
- **H2**: The proposed method raises safety of the planned access paths.
- **H3**: The proposed method reduces the planning duration.

The study is conducted in a workshop with three radiologists \( R_1, R_2 \) and \( R_3 \) from 3 different radiological departments. The experience and expertise of the 3 radiologists in the field of interventional radiology and RFA in the liver in particular is high and comparable to each other. Access paths have to be determined for RFA of liver tumors for 20 cases (6 female and 14 male patients, age: 45 -86 years) that are provided by the three departments. The acquisition parameters of the images (36-203 slices, 512 x 512 voxels in plane, slice thickness 1.5 mm - 5 mm, slice spacing 0.8 mm - 5 mm) cover the complete range of parameters that are typical for this kind of intervention. Figure 3.4 shows the results of the proposed method for one case of the study.

For comparison of the proposed method with the typical clinical workflow, the path planning is carried out twice. In the first session, no visualization support is used. The second session takes place the next day and utilizes the proposed method. In this session the cases are presented in a different order to reduce memory effects. For an efficient evaluation a questionnaire is integrated into the software prototype.

**Assessment of the support offered by the visualization (H1)** To investigate hypothesis \( H1 \), the radiologists have to fill in a questionnaire for each case during both sessions. To be able to judge the difficulty of the cases and for a subsequent comparison
3.3. Display of Infeasible Shaft Points in 2D Slice-Based Visualizations

Figure 3.4: Definition of the target point (*red square in the center of the tumor mask*) leads to immediate highlighting of points that belong to infeasible paths (*left*). In this case, only paths from left anterior directions are possible in the slice of the target point. An according path is only selected by $R3$. $R1$ and $R2$ consider these paths as unsuited and choose skin entry points that are 8 and 12 slices below, respectively. The path defined by $R2$ is displayed on the right side. The shaft point could be found easily due to the “hole” in the visualization (*upper right*). All points that are not red belong to paths that do not penetrate any of the segmented risk structures. Slicing reveals that this hole corresponds to a clearance between lung and rib (*lower right*).

of the paths, they specify whether a double oblique path is necessary or not. To capture the subjective appraisal of the effect of the visualization, the radiologists have to specify, whether the visualization influences their decision regarding the choice of the access path (Table 3.1, column 2) and if a path is chosen that would not be found without the visualization (Table 3.1, column 3). All radiologists are influenced by the visualization in a considerable number of cases (55% on average). However, not for all of these cases the path chosen in the second session differs from the path chosen in the first session. The radiologist might have found the same path but easier or faster. Even for some cases for which a radiologist states that a path is chosen that would not be found without the visualization the path of the second session does not differ significantly from the path of the first session.

Additionally, the radiologists have to assign a mark denoting how much the visualization supports them in finding a suited access path (Table 3.1, column 4). Mark 1 is assigned if the visualization is of extraordinary help and mark 6 for cases
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Influenced decision

<table>
<thead>
<tr>
<th></th>
<th>New path chosen</th>
<th>Average mark (all cases)</th>
<th>Average mark (difficult cases)</th>
</tr>
</thead>
<tbody>
<tr>
<td>R1</td>
<td>13 (65 %)</td>
<td>11 (55 %)</td>
<td>2.35</td>
</tr>
<tr>
<td>R2</td>
<td>12 (60 %)</td>
<td>0 (0 %)</td>
<td>2.25</td>
</tr>
<tr>
<td>R3</td>
<td>8 (40 %)</td>
<td>1 (5 %)</td>
<td>3.7</td>
</tr>
<tr>
<td>Average</td>
<td>55 %</td>
<td>20 %</td>
<td>2.77</td>
</tr>
</tbody>
</table>

Table 3.1: Results of the questionnaire regarding the support offered by the visualization in session 2.

in which the visualization is of no help at all. It can be noticed that the average marks for the visualization support are good but not excellent (R1: 2.35, R2: 2.24, R3: 3.7). The reason is obvious: the visualization support is rated poorly almost always for easy cases (i.e. cases that allowed for in-plane access paths). For such cases no additional visualization is required since the radiologist is able to find a suited path immediately without any software support. Hence, a more meaningful assessment can be ascertained if only complicated cases are taken into account. If the assessment is restricted to cases that require a double oblique access path the average marks assigned by the three radiologists are 1.62, 2.17, and 2.88 (Table 3.1, column 5).

Influence on safety of the planned paths (H2) To verify hypothesis H2, the paths stored during both sessions are analyzed with respect to penetration or grazing of risk structures. During session 1, seven access paths are selected that either penetrate or graze a rib or the lung. In session 2 this does not happen at all. However, one of these paths, which has been planned by R2, penetrates a part of the ribcage that is not correctly segmented. Hence, penetration would not have been detected by the proposed visualization either. The remaining six cases were planned by R1. In three of these cases, the path selected in session 2 is similar to the one selected in session 1. In the three other cases, a different intercostal space is selected in session 2.

For case #16, R1 chooses an infeasible path 1 in session 1 which penetrates a rib (Figure 3.5 (c)). Using the proposed method, R1 selects a very similar path 2 in session 2 which however does not violate any risk structure (see Figure 3.5 (b, d, f)). In Figure 3.5 (f) the placement of the shaft point of path 2 in a region not labeled as infeasible by the visualization can be seen. Figure 3.5 (e) shows the proposed visualization for the shaft point slice of path 1, although it is not used in session 1 in order to show that the visualization would have predicted that this path is not valid.

Case #9 is among the cases for which the paths selected in session 1 and 2 are very different. In session 1, R1 selects a path with a shaft point in superior direction of the target point (therefore, the path points ”upward” starting from the target point). It penetrates the lung and a rib (Figure 3.6 (c)). The path defined in session 2 on the other hand points ”downward” (shaft point in inferior direction of the target...
3.3. Display of Infeasible Shaft Points in 2D Slice-Based Visualizations

Figure 3.5: Case # 16: Path 1 from session 1 (a, c, e) and path 2 from session 2 (b, d, f) are very similar. However, path 1 penetrates a rib (red arrow).
Figure 3.6: Case #9: Path 1 \((a, c, e)\) is completely different to path 2 \((b, d, f)\) and penetrates the lung and one rib (red arrow). Note: the horizontal red line visualizes the position of the coronal reformation in the software and is not part of the visualization.
point) and does not penetrate a risk structure, although it is relatively close to a rib (path 2, Figure 3.6 (d)). The according visualization results can be seen in Figure 3.6 (e, f). Note, that for Figure 3.6 (e) not the slice of the shaft point is chosen, as this was placed inside the patient, where the visualization would not be visible. Instead, the slice in which the chosen path penetrates the skin is displayed. It can be seen that the instrument shaft would pass through a red region. Therefore, the collision would have been predicted if the visualization was used in session 1.

**Influence on planning duration (H3)** To be able to assess the influence of the proposed visualization method on the planning duration, the time needed for path definition is measured in both sessions. The measurement is started once the placement tool is activated and stopped as soon as the radiologists start filling in the evaluation questionnaire. An overview of the time measurements is given in Figure 3.7. A considerable reduction of the planning duration can be noticed for radiologists R2 and R3: the mean duration in session 2 was 57.7 s and 22 s compared to 124.6 s and 43.6 s in session 1, which equals to a reduction to 54% and 50%, respectively. However, a slight increase of 14% from a mean duration of 33 s (session 1) to 37.7 s (session 2) can be observed for radiologist R1.

**Interviews** Finally, several interviews are conducted with the radiologists. All of them state that the method is of high value for the planning of complicated cases. More precisely, it greatly supports the selection of a suited intercostal space during the determination of double oblique access paths. Although it is not needed at all for

![Figure 3.7: Durations measured for the radiologists R1, R2 and R3 during session 1 (S1) and session 2 (S2). Durations given in seconds.](image)
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easy cases, the radiologists do not see it as a distraction. If the visualization should indeed be a burden, it could be hidden very easily by moving the mouse pointer to a point inside the patient or outside the viewer. Hence, it could be used as a standard tool. Furthermore, the visualization has the most impact when used in the axial view. The other views are only used to judge the degree of angulation of the selected paths.

3.3.5 Discussion

The evaluation of the presented method was carried out on the basis of incomplete segmentation results. Notably, segmentation masks of hepatic and epigastric vessels were missing. Even with this limitation, the presented method offers a significant support during the planning of hepatic access paths since the initial selection of a path can be accelerated significantly. Although the clinicians, who carried out the study, encouraged the inclusion of further risk structures, they did not judge the omission of these structures as a fundamental restriction of the method. The speed-ups as well as the marks that have been gathered in the study exhibit strong variations although all radiologist share similar experience. It can also be observed that good marks do not correlate with good speed-ups. One possible cause might be the very different background of the three radiologists. The RFA workflow varies strongly between the three radiological departments (MR-guided vs. CT-guided vs. CT-fluoroscopy). Hence, the educational background can be assumed to be different among the radiologists and might have influenced the evaluation. Radiologist R2 is familiar with 3D visualizations and might have a slight preference for such techniques. Additionally, it was noticed in advance that Radiologist R3 had certain reservations against the proposed method because he didn’t think that it would be of high relevance for him due to his experience. Both facts seem to be reflected in the marks. But still a significant speed-up could be observed for both testees. The discrepancy between the good marks of Radiologist R1 and the observed slowing down might be related to the fact that in 11 out of 20 cases he chose a completely different path in session 2 (Table 3.1, column 3). It seems that he used the visualization technique to explore various solutions. Despite the observed discrepancies, the evaluation and the discussions with the clinical partners showed, that the proposed 2D visualization method is suited to augment the typical clinical workflow. The most important attribute is that the visualization still allows inspecting the original intensity values of the CT or MR scan. The radiologists can explore the image data in exactly the same way as they learned it. Additionally, the interaction required to slice through the image stack is much easier than rotating a 3D rendering. This is also true for the definition of the virtual applicator, even if only the shaft has to be moved. However, it could be argued that a 3D visualization might offer a better overview of the anatomical situation. Hence, for physicians, who are familiar with 3D renderings, the benefit of a method purely enhancing the 2D visualization might be marginal. Therefore, investigation of the translation of the method for three-dimensional visualization
might be meaningful. Also, the 3D visualization might allow better judgment of the distance of the selected path to the risk structures than the 2D visualization alone. However, this could also be improved in both, 2D and 3D visualizations, by incorporating an additional safety margin.

The 2D visualization highlights all infeasible points outside of the patient. As a result, the user is not restricted to the selection of a skin entry point. Instead, an arbitrary shaft point outside the patient body can be selected. This enables the modification of the angle between the trajectory and the axial plane in a certain range without the need to select another slice. Furthermore, this implementation of the visualization allows the user to judge the distance of the selected path to risk structures in proximity directly in the visualization. However, it was observed that this visualization may be confusing at first to some users since the perspectively projected risk structures result in shapes on the 2D slice which cannot be easily interpreted. An alternative masking of the visualization to the patient contour might reduce the complexity of the visualization.

A minor inaccuracy is introduced due to the modeling of the instrument shaft as a line with no thickness. However, depending on the used instrument, the shaft has a diameter of more than 2 mm. The instruments of the Medtronic/Covidien Emprint™ MWA systems for example have a shaft diameter of 2.4 mm. To raise the accuracy of the method, and therefore the safety of the planning process, the shaft thickness should be incorporated.

The visualization presented in this section is restricted to pixels that are inside the original image slices. However, depending on the field of view used during image generation, the patient contour might not be completely covered. As a result, the visualization would not be visible adjacent to such sections (see Figure 3.6 (e)). The visualization should be adapted such that the visualization is visible on the whole circumference.

A serious limitation of the method is that only possible collisions between the target point and the shaft point are taken into account. The instrument is modeled as a line starting at the target point and extending through the shaft point (Figure 3.8 (a)). For an application to biopsy planning, this would pose no restriction, as the target point would be close to the tip. However, in the case of ablation procedures the target point is not situated at the instrument tip (Figure 3.1). Hence, the instrument extends into the opposite direction and might collide with risk structures "behind" the target (Figure 3.8 (b)). An obvious solution would be to use the instrument tip as target point (Figure 3.9 (c, d)). However, for interaction with the instrument it is much more convenient to use the center of the ablation zone as the target point. That way, rotation around the target point does not move away the expected ablation zone from the target volume (Figure 3.9 (a, b)). Hence, to allow for convenient interaction and for safe planning of cases in which the tumor is close to risk structures, the method should be adapted such that structures behind the target are integrated into the projection as well (Figure 3.8 (c)). Furthermore, it should be investigated if the method can be translated to the definition of the target point, which has been assumed to be fixed so far. This might be of
Figure 3.8: Projection of risk structures from the instrument tip correctly predicts risk structure penetration (a). The candidate shaft points 1 and 2 would not define infeasible access paths. However, if the target point is located away from the tip, collisions of the tip with the vessels 1 and 3 are not considered (b). For a correct prediction, the projection of these vessels would have to be propagated to the opposite side (c).

Figure 3.9: Using the center of the ablation zone as target point allows for convenient rotation of the instrument (a and b). If the tip is used instead, the ablation zone is moved away from the target volume (c and d).
interest especially for planning of ablation of larger tumors, which requires multiple ablations at various target points. For this scenario it would also be required that other existing instruments are taken into account similar to risk structures. Finally, the integration of additional planning criteria should be investigated. Especially the incorporation of strict criteria, which cannot be violated in any circumstance, such as the maximal path length for example, should allow for an enhancement of the planning support without raising the visual complexity.

3.4 Display of Infeasible Access Paths in 2D- and 3D-Visualizations

The method presented in the previous section shows potential to improve the established planning workflow. However, several shortcomings and possible improvements have already been identified (see Section 3.3.5). Hence, this section aims at extension of the previous method with regard to the following aspects:

- Consideration of the shaft thickness
- Incorporation of the instrument tip
- Incorporation of safety margin
- Incorporation of other instruments as obstacles
- Translation to the definition of the target point
- Restriction of path length
- Improved 2D visualization
- 3D visualization

The two stages of map generation and visualization are maintained. As before, the visualization has to project the generated map and mask it in order to reduce visual occlusion. However, to adapt the method according to the requirements above, an additional stage is inserted at the beginning: the obstacle preparation stage. While the risk structure mask is directly used for the risk structure map generation presented in Section 3.3.1, derived representations are used here to achieve the defined goals. Since other instruments are incorporated as well, the term obstacle is used to refer to structures that must not be penetrated by a valid path. Consequently, the term obstacle map is used for the modified risk structure map concept presented here. Various of these maps are introduced to achieve the goals defined above. In the following, solutions for each of the mentioned requirements are presented. Each of these might require modifications to one or several of the three stages. Figure 3.10 gives an overview of the stages of the method.
3.4.1 Consideration of the Shaft Thickness

The method presented in Section 3.3 assumes that the instrument can be modeled as a line (see Figure 3.11 (a)). However, that way a shaft point that is not in a red sector could still define an infeasible path since the real instrument with a shaft radius \( r_s > 0 \) might penetrate a risk structure (see Figure 3.11 (b)).

To solve this problem, the target point obstacle map \( M_t \) is introduced. Similar to the previous risk structure map it is generated using the target point as projection center and stores information used for penetration detection during evaluation of candidate shaft points.

The incorporation of the shaft thickness is realized in the obstacle preparation phase by modifying the obstacle geometry which is rendered into \( M_t \). As illustrated in Figure 3.11 (c), the collision of an instrument shaft with the radius \( r_s \) can correctly be predicted by offsetting the original risk structure surfaces along their surface normals by \( r_s \) before the actual obstacle map generation. The red sectors resulting from projection grow accordingly and moving the shaft point slightly outside the red sector defines an instrument placement without collision of the cylindrical shaft with any segmented risk structure (see Figure 3.11 (d)).

The risk structures are rendered into \( M_t \) utilizing DVR. To implement the proposed surface offset, an Euclidean distance transform (DTF) [49, p.270] based on the risk structure mask is computed in a preprocessing step. The result of this computation is a 3D image that stores for each voxel the distance to the closest point that belongs to the mask and is called risk structure DTF in the following. The offset surface is rendered by means of DVR of the risk structure DTF with an adequate transfer function (see Section 2.4.1). To this end, a step function is chosen which maps all values less than or equal to certain threshold to an opaque white and makes all other values completely transparent. This transfer function threshold, is set to \( r_s \) in this case.

Alternatively, morphological dilation could be used to create the surface offset. However, this computation would have to be updated each time \( r_s \) changes due to the usage of a different instrument which can lead to a noticeable delay. The solution based on the DTF and a transfer function only requires an update of the transfer function and \( M_t \).
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

Figure 3.11: The risk structure map predicts collision correctly for instrument shafts with no thickness (a). However, a realistic instrument has a cylindrical shaft with the radius $r_s$. Hence, a path marked as valid by the projection might still lead to a collision for such instruments (b). The collision can be predicted correctly by offsetting the risk structure surfaces by $r_s$ (c and d).

3.4.2 Incorporation of the Instrument Tip

As discussed in Section 3.3.5, the method presented in Section 3.3 only considers risk structures that would be penetrated by the section of the shaft between the target point and the handle. Collisions of the section between the target point and the tip of the instrument are not considered. In the following, this whole section of the instrument will simply be referred to as instrument tip. To detect collisions of the instrument tip with obstacles, the cube map concept can be utilized as well. As illustrated in Figure 3.8 (c), this mainly involves checking for collisions in the opposite direction with regard to the target point. Hence, using an inverted lookup vector $v'$ penetrations of obstacles by the instrument tip can be detected (see Figure 3.12). However, only obstacle parts that are close enough to the target point should be considered. More precisely, a point $p$ inside an obstacle is only taken into account if its distance $d_p$ to the target point is smaller or equal to the length of the instrument tip $l$. Furthermore, the shaft radius $r_s$ has to be subtracted to compensate for the offset introduced in Section 3.4.1. Hence, a point $p$ is considered an obstacle if:

$$d_p \leq l - r_s.$$ (3.1)

Consequently, the depth must be evaluated during the visualization stage. To this end, the target point obstacle map $M_t$ is modified such that the distances of obstacles to the target point are stored instead of the previous binary information (see Section 3.3.1). This is inspired by the z-buffer as described in Section 2.4.3 and its usage for shadow mapping. However, in this case, the depth values are linearly normalized.
using a shader. Depth values are mapped from the range $[0, d_f]$ to $[0, 1]$ where the 
*far distance* $d_f$ is set to the shaft length of the used instrument. Figure 3.13 shows an exemplary target point obstacle map containing normalized depth values.

During projection of $M_t$ in the visualization stage, absolute depth values are reconstructed based on $d_f$. Algorithm 2 outlines the shader used for projection. Note, that the absolute depth values are only necessary for the collision detection of the instrument tip. They are not needed for the detection of penetrations by the shaft as it can be assumed that the shaft ends (and the handle begins) outside the body, while all risk structures are inside the body. Hence, it is sufficient to check if any risk structure would be penetrated by an infinite shaft (in direction $v$ in Figure 3.12). This is also the reason why depth values were not needed at all for the method presented in Section 3.3.

Figure 3.13: Obstacle map for a target point in the liver dome. The large dark regions belong to the lung surface, which is closer than the rib cage. Hence, the relative depth values are smaller. White pixels correspond to directions with no penetration. Note, that vascular structures are not included here for reasons of clearness.
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

Algorithm 2 Shader pseudo code for determination of the validity of the potential shaft point \( s \) for the given target point \( t \) and the corresponding target point obstacle map \( M_t \), which contains normalized distances of obstacles to \( t \). The function \( \text{absDepth()} \) is used to compute the absolute depth based on the far distance \( d_f \). For the direction \( v \) it is sufficient to check if any object is on the trajectory \((d < 1)\).

\[
\text{function } \text{IsInfeasiblePoint}(s, t, M_t, d_f, l, r_s) \\
\begin{align*}
    v &\leftarrow s - t \\
    v' &\leftarrow -v \\
    d &\leftarrow \text{cubeMapLookup}(M_t, v) \quad \triangleright \text{Get depth for direction } v \\
    d' &\leftarrow \text{cubeMapLookup}(M_t, v') \quad \triangleright \text{Get depth for direction } v' \\
    c &\leftarrow (d < 1) \quad \triangleright \text{Shaft would collide} \\
    c' &\leftarrow (\text{absDepth}(d', d_f) \leq (l - r_s)) \quad \triangleright \text{Tip would collide} \\
\end{align*}
\]

return \((c \text{ or } c')\)

3.4.3 Incorporation of Safety Margin

The visualization presented in Section 3.3 segments the space into cones that contain directions that are feasible or not feasible. A distinction based on the distance to the risk structures is not made. However, due to possible placement inaccuracies and differences between the patient anatomy during planning and during procedure, paths with a certain distance to risk structures are preferred. To allow for an according distinction, a safety margin is introduced. Conceptually, this margin can be either added to the instrument or to the obstacles in all directions. Using the latter approach, the problem is analogue to the incorporation of the instrument thickness (Section 3.4.1) and can be solved by the previously described technique for offset surface generation. However, to allow for a distinction of the safety margin and infeasible paths during visualization, the obstacle representation is not changed for the generation of the target point obstacle map \( M_t \). Instead, an additional, modified target point obstacle margin map \( M_t^+ \) is introduced. It is generated by rendering the risk structure DTF with a modified transfer function threshold \( d = r_s + m \) where \( m \) is the desired safety margin.

In the visualization phase, \( M_t^+ \) is projected in the same manner as the margin-less versions \( M_t \). The depth evaluation is identical. The output color is computed based on both projection results:

\[
c(s) = \begin{cases} 
(1, 0, 0, 1) & \text{if } \text{IsInfeasiblePoint}(M_t, s) \\
(1, 1, 0, 1) & \text{if } \text{IsInfeasiblePoint}(M_t^+, s) \text{ and not } \text{IsInfeasiblePoint}(M_t, p) \\
(0, 0, 0, 0) & \text{else} 
\end{cases}
\]

(3.2)

Hence, if a penetration is detected in \( M_t \), the potential shaft point \( p \) is displayed in red. If a penetration is detected in \( M_t^+ \) but not in \( M_t \), yellow is used.
3.4.4 Incorporation of Other Instruments as Obstacles

For larger target volumes, multiple instruments might be used to ablate the tissue. And, depending on the used modality and instrument model, these instruments might be in the patient at the same time. In that case, possible collisions inside and outside the patient have to be taken into account during planning already. The obstacle preparation phase is adapted accordingly by representing all instruments besides the one that is currently being modified as obstacles. These instruments are called *obstacle instruments* in the following. For the representation of the obstacle instruments polygonal representations are used since the instruments are available in that form in the planning application for visualization of the plan in 2D and 3D visualizations. Furthermore, this representation allows for creation of offset surfaces without the need of DTF computation, which would have to be updated after each change of the obstacle instruments. Offset surfaces are needed for the same reason as described in Section 3.4.1: to compensate the missing thickness of the shaft of the instrument that is currently manipulated as this instrument is modeled as a line. Hence, the shaft surface of each obstacle instrument is also offset to the outside by the shaft radius \( r_s \), or by \( r_s + m \) in the case of the target point obstacle margin map \( M^+_t \) (Figure 3.14 (a)). The offset is implemented using a vertex shader. It moves all vertices that are not on the main shaft axis perpendicular to that axis, and the point at the tip along the axis.

The handles of the obstacle instruments are treated in a very similar way. However,

![Diagram](image)

Figure 3.14: Modification of obstacle instrument geometries in the obstacle preparation phase: vertices of the shaft surfaces are offset by the shaft radius \( r_s \), or, in the case of margin map generation, by the shaft radius \( r_s \) and the margin distance \( m \) (a). For the offset of the handle surfaces, the handle radius \( r_h \) is used instead of \( r_s \) (b). As a result, the collision of the handle of the currently modified instrument with an obstacle instrument (c) can be approximated by checking the collision of a line with the displaced obstacle instrument (d).
outside the body the handle of the manipulated instrument would collide with the handle of the other instruments. Hence, the missing handle thickness of the currently modified instrument has to be compensated instead of the shaft. To achieve this, the handle radius $r_h$ is used accordingly as offset distance (Figure 3.14 (b)). The vertex shader simply moves all vertices of the handles along their surface normal. Figure 3.14 (d) illustrates the approximative detection of the collision of two instrument handles.

### 3.4.5 Application to Target Point Definition

Since the physician only manipulates the target point or the shaft point at a time, the path definition problem can be treated in a symmetric manner:

1. During the manipulation of the shaft point, all possible candidate instrument placements sharing the same fixed target point are evaluated. The according visualization is called **shaft point evaluation visualization**.

2. During the manipulation of the target point, all possible candidate instrument placements sharing the same fixed shaft point are evaluated. The according visualization is called **target point evaluation visualization**.

So far, only the first case has been investigated. However, for large target volumes the manual modification of the target point is of similar interest. Hence, the **shaft point obstacle map** $M_s$ and **shaft point obstacle margin map** $M_s^+$ are introduced. The computation is equivalent to that of $M_t$ and $M_t^+$, respectively. Neither obstacle representation nor the map generation have to be changed besides the modified projection center which is now set to the given shaft point. The visualization stage however is different with regard to masking and depth evaluation. The target point can be placed everywhere in the target volume. Hence, the visualization is restricted to the target volume mask instead of voxels outside the patient.

The modified depth evaluation considers the length of the instrument tip $l$ and the distance $d_{st}$ between the given shaft point $s$ and any potential target point $t$ (see Figure 3.15). The latter is infeasible, if the instrument tip collides with an obstacle.

Figure 3.15: Illustration of the depth evaluation for the fixed shaft point $s$ and the candidate target point $t$: a collision of the path defined by these two points with any obstacle occurs if the distance of an obstacle to $s$ is smaller than the distance of the tip to $s$, which is given by $d_{st} + l$. 
This is the case, if the distance of the obstacle to $s$ is equal to or smaller than $d_{st} + l$.

The according adaption of the shader outline in Algorithm 2 for the evaluation of each potential target point $t$ is given in Algorithm 3. The evaluation of the shaft point obstacle margin map $M^+_t$ is equivalent. The combined visualization of both results is carried out as described for $M_t$ and $M^+_t$ before (see Section 3.4.3).

**Algorithm 3** Shader pseudo code for determination of the validity of the potential target point $t$ for the given shaft point $s$ and the corresponding shaft point obstacle map $M_s$, which contains normalized distances of obstacles to $s$.

```plaintext
function IsInfeasiblePoint($t, s, M_s, d_f, l, r_s$)
    $v ← t - s$
    $v' ← -v$
    $d ← cubeMapLookup(M_s, v)$  // Get depth for direction $v$
    $d' ← cubeMapLookup(M_s, v')$  // Get depth for direction $v'$
    $c ← (absDepth(d, d_f) ≤ (length(v) + l - r_s))$  // Tip would collide
    $c' ← (d' < 1)$  // Handle would collide
    return ($c$ or $c'$)
```

### 3.4.6 Restriction of Path Length

Among the planning criteria presented in Section 2.2.2, there are two criteria which render an access path infeasible if violated: first, under no circumstances, a risk structure should be penetrated. Second, paths, that are longer than the available instrument, can not be implemented. The other criteria influence the suitability, but not the feasibility. The proposed method considers the first of the two mentioned criteria. The integration of the second is possible without compromising the simplicity of the visualization. To this end, the existing shaders are marginally changed. As the target point of an instrument is fixed on the shaft of the instrument, the maximum distance between the target point and any shaft point is limited by the distance $d_h$ of the target point from the handle, which is a instrument model specific constant. Hence, the shaders are modified as follows:

- During shaft point evaluation visualization, all points, which are farther away from the fixed target point then $d_h$, are displayed as infeasible.

- During target point evaluation visualization, all points, which are farther away from the fixed shaft point then $d_h$, are displayed as infeasible.

Hence, a path is visualized as being infeasible, if it either penetrates a risk structure, or its length exceeds $d_h$. The visualization of the violation of the path length is subject to the same masking as the visualization of the obstacle penetration. Furthermore, the safety margin $m$ can also be applied to the penetration depth evaluation by comparing the distances with $d^+_h = d_h - m$. 

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3.4.7 Improved 2D Slice Visualization

As discussed in Section 3.3.5, the masking aspect of the presented 2D visualization (Section 3.3.2) should be modified to fulfill the following requirements:

1. Enable visualization outside original image if the patient is not contained completely inside.
2. Restrict visualization to contour around the patient body.

The first requirement is accounted for by increasing the patient mask image size in the $x$ and $y$ direction according to the desired boundary thickness, and by extending the rectangle onto which the shader is applied accordingly (Section 3.3.2).

To modify the masking such that the visualization is restricted to the body contour, image processing is used. In principle, morphologic operations can be used to create a border around the patient mask. However, that way, the display of the boundary would suffer from strong aliasing artifacts (see Figure 3.16 (b)). Instead, a Euclidean DTF [49, p.270] is used to compute the distance of every voxel to the closest point that belongs to the patient. This patient mask DTF is provided to the shader which thresholds it during rendering. The result is used to mask the projection of the obstacle maps (see Figure 3.16 (c)). This approach also allows to change the thickness of the contour without the need to execute the image processing again.

3.4.8 3D Visualization

The proposed visualization methods can be translated easily to 3D visualizations at least for the definition of the shaft point since this point is placed on the skin which is clearly visible in such a visualization and not occluded by other structures. Without any modification, the target point obstacle map $M_t$ and the target point obstacle margin maps $M'_t$ can be projected onto the skin using the same shaders.

Figure 3.16: Comparison of various approaches to mask the shaft point evaluation visualization: unrestricted display of path suitability outside the patient mask (a), restricted to body contour based on binary mask (b) and utilizing a DTF (c).
as in the 2D visualization, assuming that the skin is visualized by means of surface rendering\(^1\). To this end, a mesh representing the skin is extracted from the patient mask DTF (Section 3.4.7). The visualization is further modified to allow for convenient selection of in-plane access paths. To this end, the shader additionally displays a contour highlighting the intersection of the skin with the transverse plane of the slice in which the target point is situated. Furthermore, the skin entry point is highlighted (see Figure 3.17).

### 3.4.9 Implementation

The proposed visualization method is implemented using the rapid prototyping platform MeVisLab [164, 139], which uses Open Inventor\(^\text{TM}\), an object-oriented 3D toolkit [178], as the basis for visualization. The method is integrated in the Software Assistant for Interventional Radiology - SAFIR [200, 162] (Figure 3.17), which provides interactive methods for segmentation of tumors [140] and vessels [12] in CT images. The patient, ribs and lungs are automatically segmented, mainly based on thresholding.

\(^1\)See also Section 2.4.1. Here, the proprietary neighboring cells algorithm of MeVisLab [139], which is similar to the well known marching cubes algorithm [154, p.162], is used.

![Figure 3.17: Visualization of infeasible zones and the safety margin on the patient skin in a 3D rendering inside SAFIR. The additional blue line in the transverse plane of the target point allows for intuitive improvement of path 1 and the definition of an in-plane access path 2.](image-url)
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

The evaluation of the cube map during rendering is supported by modern GPUs [199] and exposed through shading languages. The implementation uses GLSL (Section 2.4.2), which is supported in MeVisLab by dedicated shader frameworks for various aspects like volume rendering [160], surface rendering, 2D view overlays and post rendering effects. To fully exploit the theoretical potential of the cube map with respect to performance, it is of great importance to update a cube map only when necessary. Otherwise, the achievable frame rates are reduced. As the proposed method uses four obstacle maps, an update of all these maps on every redraw could, depending on the used hardware, result in unusable frame rates. Hence, each obstacle map is updated only when required by a certain interaction or the change of relevant input data or parameters. The achieved frame rates are reported in Section 3.4.10.

The visibility of the shaft point and target point evaluation visualizations is toggled in a context sensitive manner depending on the current interaction. Each visualization is turned visible as soon as the mouse pointer hovers over the corresponding interaction widget and turned invisible as soon as the mouse pointer is moved away from it. In the 2D views, the widgets are simple small squares, which are used to visualize and modify the shaft and target points. In the 3D rendering, the handle of the selected instrument is used to change the instrument orientation. Consequently, the visualization is visible the whole time the user modifies the instrument, which is achieved by clicking and dragging the described interaction widgets.

A special case is the creation of a new instrument, for which the user first places a target point and then the shaft point. In this case, the shaft point evaluation visualization is turned visible as soon as the user places the target point. It stays visible until the user has placed a shaft point. That way, the shaft point evaluation visualization can be used even though there is no shaft point widget to hover over yet.

Finally, the visualization is only shown in the viewer in which the user interacts. That way, the occlusion of the original image data is minimized. Figure 3.18 outlines the overall method including the described update and visibility dependencies.

3.4.10 Results

The described visualization method is seamlessly integrated into the axial 2D views, the 3D view and the instrument-aligned MPR views of SAFIR. The computational overhead for obstacle scene preparation and obstacle map generation is minimal. A slight delay can be noticed for example when the instrument selection is changed, since this leads to an update of both stages. However, it is below one second and does not disturb the user experience. The visualization of the maps marginally influence frame rates during instrument manipulation. On a computer with an Intel® Core™i7 940 and a NVIDIA GeForce® GTX 670, for example, the frame rate is reduced from 57 frames per second (fps) to 55 fps.

Figure 3.19 shows the shaft point evaluation visualization for a simple in-plane setting. It illustrates the modified masking, the yellow zones resulting from the projection of the obstacles with according margins (a safety margin of 3 mm was
Figure 3.18: Stages and update dependencies (red, dashed lines) of the proposed visualization method: obstacle representations are created in the obstacle scene preparation phase based on a risk structure mask, obstacle instruments (all instruments except the one currently manipulated) and the user definable safety margin. Changes to any of these inputs trigger an update of this phase. The generation of the obstacle maps is updated, when the respective projection center has been changed and after each update of the obstacle scene (not displayed here). The visualization is automatically updated by the event handling of Open Inventor™ during redraws. The visibility of the three visualizations (shaft point evaluation visualization in 2D and 3D and the target point evaluation visualization) depends on the current instrument interaction (green, dotted lines).
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

Figure 3.19: Shaft point evaluation visualization: the initially placed target point (red square, black arrow) does not allow for a shaft point that does not violate the safety margin (a). For the displayed placement, the instrument tip is very close to a vessel (yellow arrow). Rotating the instrument clockwise around the target point would move the tip outside of the safety margin of the vessel, but the shaft would be moved towards the rib. (b): Counter-clockwise rotation lets the tip collide with the vessel (red arrow) as soon as the shaft point is moved into the red zone (white arrow). As soon as the target is moved towards the target volume center, a safety corridor is opened in the visualization (c). Moving the shaft point into the red zone leads to collision with the rib (d, red arrow).

The target point evaluation visualization for the same setting is shown in Figure 3.20. The visualization, which is masked by the target volume, is combined with the target volume visualization, which is available in SAFIR. In subfigure (d), the incorporation of the shaft thickness can be seen: the target point is placed exactly on the boundary of the red zone. As a result, the shaft marginally collides with the border of the vessel (red arrow).
Figure 3.20: Target point evaluation visualization: the initially placed instrument is valid as the target point is outside any red region. Furthermore, it is outside the safety margin (a). The large red region in the target volume corresponds to target points that would define an infeasible path because the instrument tip would collide with a vessel behind the target volume (b, red arrow). Moving the target point towards the target volume center onto the boundary of the yellow zone leads to a clearing between tip and the vessel that corresponds to the chosen safety margin (c). The small red region parallel to the course of the instruments corresponds to target points that would lead to a collision of the shaft with a vessel between target volume and skin (d, red arrow).
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

Figure 3.21: Shaft point evaluation visualization for a slightly angulated path (slice 76; target point in slice 82): For the chosen infeasible path (a), even the instrument-aligned MPR does not show the collision initially (b). Rotation of the MPR around the instrument eventually reveals the collision with a vessel behind the target volume (c, red arrow). The visualization can also be applied to the MPR view (c).

Figure 3.21 illustrates how the visualization can help to identify infeasible angulated paths that would be not obvious in the axial views nor in an instrument-aligned MPR if the orientation around the instrument is not optimal. Although the MPR shows the route of the whole access path, the collision with a vessel is not visible until the user has rotated the MPR more than 90° around the instrument. The proposed visualization on the other hand indicated this path as infeasible even before the shaft point has been moved into the red zone. The application of the visualization to an MPR is shown in Figure 3.21 (d). The detection of collisions of the currently manipulated instrument with other, already placed instruments is shown in Figure 3.22. Although two in-plane placements have been chosen in order to allow for a good depiction of both instruments, this works as well for double-angulated placements due to the underlying projection approach.

As a second criterion influencing the feasibility of access paths, the restriction of the maximal access path length has been integrated (see Figure 3.23). The overall
Figure 3.22: Placement of two instruments: while the initial placement of the selected instrument (light gray) is valid, moving the shaft point into the red zone leads a collision of the handle with the other instrument (b). Similarly, the target point evaluation visualization predicts tip collisions (c, d).

Figure 3.23: The incorporation of the path length restricts feasible shaft points (a, red arrow). The visualization is adapted accordingly for a longer instrument (b, green arrow). Similarly, the maximum depth and the safety margin are displayed in the target during target point manipulation (c, d).
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

Visualization is not changed as this simply adds a decision criterion for the selection of the visualization color. The visualization is immediately adapted when the type of the selected instrument is changed (Figure 3.23 (b)).

Finally, the integration into SAFIR and the translation to a 3D rendering is shown in Figure 3.17. The skin mesh is slightly offset outwards using the patient mask DTF, in order to allow for a combined visualization with the DVR of SAFIR [162].

3.4.11 Evaluation

During a small workshop, the proposed visualization method is presented to two interventional radiologists $R_1$ and $R_2$ with 10 and 1 year of experience in interventional radiology, respectively. They use SAFIR to retrospectively plan MWA ablations of liver tumors. Due to the limited time available for the workshop ($\sim 2$ h per physician including discussions), only four different cases are investigated. Hence, the aim of the study is not to obtain quantitative results. It can only serve to get a first impression of the benefits of the method for the radiologists. The selection of patient image data mainly contains complicated cases, since the radiologists agreed to the hypothesis, that the method has a stronger impact on the planning of such cases. Only the fourth case can be considered being easy and allows for short in-plane access paths. The planning of multiple instruments is not considered, since this method is not a focus of the participating radiologists. Also, the developed 3D visualization is not taken into account, since both radiologists agree that the benefit of 3D visualizations in general is marginal for their work.

The planning is carried out twice for each case: once without (method 1) and once with the new visualization (method 2). For the cases #1 and #3, method 1 is used first. Cases #2 and #4 are first planned using method 2. The planning duration is measured and each radiologist has to assign a mark to the selected access path expressing the overall quality of the path (six-point Likert scale, one=very good . . . six=infeasible). The results are summarized in Table 3.2. In general, planning is carried out quicker, when a case is planned for the second time, which is not surprising given that a very limited time span passed between the two planning sessions for each case. Hence, features of the cases are memorized by the radiologists. The assigned marks are identical for the two methods in most cases, even though the chosen access paths vary almost always. The variations with regard to the access paths and the assigned marks are greater for $R_2$. In two cases (#1 and #3), the results achieved with method 2 are slightly better than with method 1. However, in both cases method 2 is used after method 1. Hence, it can be argued that the knowledge acquired before influenced the path selection.

After the planning of the four cases is completed for both methods, several aspects of the developed visualization method are discussed with each radiologist individually. Both radiologists state that the proposed approach offers a good benefit for the definition of the skin entry point. Also, the incorporation of the safety margin and of collisions of the instrument tip with risk structures is considered beneficial as it can raise safety further during planning. However, the safety margins should be
Table 3.2: Results of the informal evaluation: the first two columns define the order in which the cases are planned using the methods 1 and 2. The planning durations and resulting access path qualities are given in the third and fourth columns.

<table>
<thead>
<tr>
<th>Order</th>
<th>Case / Method</th>
<th>Planning duration (s)</th>
<th>Quality Rating (1-6)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>R1 / R2</td>
<td>R1 / R2</td>
</tr>
<tr>
<td>1</td>
<td>1 / 1</td>
<td>180 / 360</td>
<td>3 / 4</td>
</tr>
<tr>
<td>5</td>
<td>1 / 2</td>
<td>60 / 160</td>
<td>3 / 3</td>
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<tr>
<td>6</td>
<td>2 / 1</td>
<td>100 / 190</td>
<td>2 / 3</td>
</tr>
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<td>2 / 2</td>
<td>140 / 310</td>
<td>2 / 3</td>
</tr>
<tr>
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<td>3 / 1</td>
<td>240 / 180</td>
<td>2 / 3</td>
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<td>30 / 80</td>
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<td>4</td>
<td>4 / 2</td>
<td>40 / 190</td>
<td>1 / 2</td>
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Chapter 3. Interactive Feasibility Visualization for Access Path Planning

treated in a more differentiated way: for example, access paths are typically planned relatively close to the upper edge of the ribs in order to spare the intercostal nerves, which are located close to the lower edge of the ribs.

The benefit for the planning of the target point is assessed as being marginal. However, it is agreed that the real benefit can only be assessed during the planning of the ablation of larger tumors. The modified 2D visualization, which restricts the display of the infeasible areas to a small margin close to the skin, is seen as an improvement in relation to the previous implementation, which covered all voxels outside the patient. The restriction of the access path length is only discussed with R2, who judges it to be a beneficial extension of the method as it could reduce the need for manual measurements.

The potential usefulness of the method for the clinical routine is discussed along with both radiologists. They share the assessment, that the method has a larger impact for complicated cases and for less experienced physicians. R2 suggests that the method can be helpful during training. R1 stresses that the value for routine planning cannot be appraised in a study alone. Instead, regular usage of the overall planning system close to the daily work is needed for any conclusive statement. Furthermore, both radiologists emphasize, that any method aiming at supporting the access path determination process must be complemented by an automatic segmentation process that does not require tedious manual delineation of the required input structures.

Finally, the user experience of the presented method is evaluated using the User Experience Questionnaire [119]. This test determines values for six scales which describe the user experience for a certain product. For that, both radiologist have to select values from 26 initial scales which represent opposite attributes (e.g. annoying and enjoyable). The resulting, aggregated values are in the range between $-3$
3.4. Display of Infeasible Access Paths in 2D- and 3D-Visualizations

(horribly bad) and +3 (extremely good). The results for the proposed visualization method with respect to the scales attractiveness, perspicuity, efficiency, dependability, stimulation and novelty are 1.083, 1.125, 0.75, 1.875, 1.167, and 1.0 respectively. These results hint at a good usability, even though this questionnaire is aimed at a higher number of participants.

3.4.12 Discussion

The comprehensive visualization method proposed in this section is an advancement of the approach presented in Section 3.3. As such, it shares the advantages identified in the according study (Section 3.3.4). Simultaneously, it improves several of the shortcomings identified in Section 3.3.5. The geometry of the instruments is now taken into account as a whole: the local thickness of the major parts of the instrument as well as the instrument tip are considered. Hence, collisions among instruments as well as with risk structures ”behind” the target are correctly detected. The evaluation of the feasibility and safety of candidate access paths now also considers the penetration depth as well as an additional safety margin which can be interactively adapted by the user.

The visualization in 2D views has been improved by restricting it to the skin. Furthermore, the method has been translated to 3D rendering, at least for the shaft point definition. A similar visualization for target point selection guidance in 3D is however less practical, since the target point is deep inside the body and occluded by many structures. This could be reduced by only rendering a small region of the available anatomical information or by reducing occlusion with the help of according transfer functions (assuming the anatomical structures are displayed using DVR). However, even then the benefit would be limited, since the target point is placed inside the target volume and not restricted to a clearly definable surface.

In contrast to the method in Section 3.3, this advanced version has not been thoroughly evaluated yet. Preliminary qualitative results have been obtained during a limited user study which does not allow to generate quantitative data. A setup consisting of two sessions separated by a considerable time span would be needed for a real comparison of the proposed method with the established planning workflow. Especially the assessment of the usefulness of the method for the definition of the target point would be interesting. The radiologists who carried out the initial evaluation pointed out, that the target point can be fixed at a certain position close to the center of the target volume if it can be assumed, that the ablation zone is large enough to cover the target volume. If a prediction of the ablation zone including heat sink effects of vessels in proximity is considered during the planning as well, the modification of the target point might be beneficial. An according setup has not been evaluated during this limited workshop though due to time constraints.
3.5 Discussion and Conclusion

In this chapter, two visualization methods for the support of manual access path determination have been presented. An essential decision during the design of both methods was to highlight infeasible paths instead of feasible ones. A more obvious choice might be to display feasible paths in green. However, the chosen solution is safer since the responsibility for the selection of a feasible and safe path remains with the user. The methods only aim at the identification of paths that should be excluded in any case in order to limit the amount of paths that the user has to take into account. This approach also allows to apply the method even if not all relevant risk structures have been segmented. Even then, the presented method offers a significant support during the planning of hepatic access paths since the initial selection of a path can be accelerated significantly. Of course, this path still needs to be checked by the radiologist since an unsegmented structure might be penetrated. However, this case occurs only rarely since the amount of unconsidered risk structures in comparison to the included risk structures is very low. Although this was not investigated quantitatively, it was found that many paths are blocked by the bones, the lung and other air filled parts like stomach and bowel (see Figure 3.3 and Figure 3.13).

Existing visualization methods for the support of the access path definition process focus on the inspection of selected paths. Instrument-aligned MPRs allow for checking of collisions of the defined path with risk structures. Still, interaction like moving the MPR along the path or rotation around it is required to check the whole path also for collision outside the displayed plane. Risk structure violations can be detected and visualized automatically [134]. However, such approaches only allow to evaluate single access paths that have already been selected. In contrast, the methods presented here give an overview of infeasible options immediately in any visualization the user is used to.

The underlying concept utilizes approaches from the field of visibility determination and shadow computation. Similar to the methods presented here, Villard et al. [192] exploited the cube map concept (although they never use that term) before for the determination of candidate zones for automatic access path computation. However, their method uses the cube map only as an intermediate data structure and not directly for visualization. It is utilized to check for every vertex on a skin mesh, if all voxels of the target volume can be reached without penetrating organs, which are also represented as polygonal structures. The duration of that process is, without considering prior polygonization, in the range of a few minutes. The result strongly depends on the chosen polygonization and is not primarily used for visualization. The visualization approach by Khlebnikov et al. [107], which has been published at the same time as the work presented in Section 3.3, is the only comparable method: their method computes a path safety volume based on the same idea as described by Villard et al. However, the evaluation of the safety is carried out for all voxels of the data set. As a result, the computation time is in the range of 5 min. The
safety volume is visualized interactively using DVR. However, this visualization has been evaluated as not being intuitive, and a simplified 2D visualization has been derived. The methods presented here, on the other hand, focus on 2D visualization. A time-consuming preprocessing is not necessary due to an important simplification: the path definition problem is separated into the independent determination of target and shaft point. Hence, not all possible paths have to be evaluated all the time, but only the paths sharing a common shaft or target point. This allows to exploit the cube map method to its full potential. The result is a real-time visualization which does not require preprocessing of any noticeable duration.

The first version of the proposed method has been evaluated and it has been shown, that the approach is suited to accelerate the process of path definition and to make it safer at the same time. The second version has only been evaluated in an informal way (Section 3.4.11). Quantitative results are not available yet. However, it can be assumed that the benefits identified for the first version also apply to the extended method due to the shared underlying concept. Several aspects, such as the translation to target point definition and the visualization in 3D renderings, have not been evaluated at all though. Hence, further efforts for the evaluation are necessary.

**Publications** The method described in section Section 3.3 including the accompanying evaluation study was published as an abstract and presentation at the 25th Congress for Computer Assisted Radiology and Surgery (CARS) in Berlin, Germany in 2011 [26]. The full paper was published in 2012 in the International Journal of Computer Assisted Radiology and Surgery (IJCARS) [27]:


The extended, second method has not been published yet. An according manuscript is currently in preparation.
4 Multi-Objective Optimization of Access Paths for Percutaneous Tissue Ablation

All roads lead to Rome.

(Medieval idiom, ∼ 1100 A.D.)

Many aspects influence the suitability of an access path for a percutaneous ablation procedure. Evaluation of all relevant criteria for all possible paths, in order to select an optimal access, is a demanding task. In this chapter, two approaches for the computation of optimal access paths for percutaneous ablation in the liver are presented which consider multiple clinically relevant criteria. The first method utilizes GPU-based projections to compute a sorted list of proposals fully automatic in a few seconds. The second approach combines the first method with a numerical algorithm in order to allow for integration of a realistic approximation of the ablation zone and for interactive, criteria-driven exploration of all possible solutions.
4.1 Introduction

All percutaneous ablation procedures face similar challenges during planning of the intervention: A linear path to the target has to be found that fulfills multiple clinical criteria. These include several geometric properties that affect the safety and practicability of the path as well as the achievable therapy outcome. Among others, the following clinical criteria have to be considered during the planning of access paths for percutaneous intervention in the liver (see Section 2.2.2):

- **Penetration depth**: the path length has to be restricted according to the length of the instrument shaft. Furthermore, shorter paths are favored.

- **Distance to risk structures**: penetration of critical structures has to be prevented. In addition, larger distances to these structures are preferred.

- **Angle to the transverse plane**: low angles between the path and the transverse plane are preferred.

- **Angle to the y-axis in the transverse plane**: inappropriate directions in the transverse plane should be prevented.

- **Probable therapy outcome**: a favorable therapy outcome, i.e. good coverage of the target volume by the ablation zone, should be facilitated.

- **Transhepatic route**: guarantee a segment of healthy liver tissue along the path.

- **Liver capsule incision angle**: prevent low angles between the path and the liver capsule.

Assessing all these criteria for all possible trajectories by visual examination of the patient anatomy alone is a demanding task and requires considerable experience. The visualization methods presented in the previous chapter represent one possible approach for computer assistance to support this process. However, only the first two of the criteria mentioned above are considered by these methods. Although the integration of additional criteria is possible, the clearness of the visualization would suffer as the complexity of the visualized information would be raised. Furthermore, the presented visualization methods aim at the visualization of binary information only. However, only for the first two planning criteria strict thresholds exist which allow to treat them in a binary manner. Integration of the remaining criteria would require the simultaneous display of multiple continuous values. An alternative way to deal with the complexity of this planning problem is the utilization of optimization methodology. As the described planning problem resembles a multi-criteria decision making problem, multi-objective optimization (MOO) is suited to support solving it. In this chapter, two novel approaches for automatic and semi-automatic access path determination based on MOO are presented. Contributions of this work include:
4.2 Related Work

- A brute-force method for automatic computation of a sorted list of optimal proposals for a given target point under consideration of multiple geometrical criteria (see Section 4.4). This method utilizes GPU-based projections and image processing to achieve a computation duration of a few seconds.

- A semi-automatic access path determination method based on image processing and numerical optimization (Section 4.5). It utilizes the previous algorithm to compute starting points for a numerical optimization scheme. During optimization, the method considers multiple criteria, most notably the coverage of the target volume by the ablation zone. For that, a fast and realistic GPU-based approximation of the ablation zone including cooling effects from nearby blood vessels is utilized. Interactive exploration of the optimization results is facilitated.

- Evaluation of the proposed methods. The results of the methods are compared to access paths manually planned by experts from the field.

The structure of this chapter is as follows: Section 4.2 summarizes prior work regarding optimization for the determination of suited linear access paths. The general optimization problem is formulated in Section 4.3. Section 4.4 and Section 4.5 present the two optimization methods. Finally, these methods are discussed and compared to each other (Section 4.6) and possible extensions of the methodology are outlined (Section 4.7).

4.2 Related Work

A general introduction to optimization can be found in Section 2.5. Here, works that utilize optimization for the support of the intervention planning process are presented.

One goal of ablation procedure planning is the optimization of the coverage of the target volume by the expected ablation zone. To this end, the ablation zone can be numerically simulated. According methods model the energy deposition, the diffusion of the heat that results from that energy deposition, and the induced damage in the tissue [145]. For thermal ablation procedures, mainly the modeling of the energy deposition varies. The heat diffusion is typically modeled using the bio-heat equation [148]. A thermal dose model or temperature thresholds according to cytotoxic levels might be used to determine the tissue damage (see Section 2.1.3). For the simulation of LITT, the energy deposition into the tissue is determined by the modeling of photon emission, which can be achieved using Monte-Carlo simulation [165, 124]. Numerical simulation of RFA typically comprises modeling of the electric field through the electrostatic equation. Kröger et al. [117] propose an according nonlinear finite element model which takes into account material parameters that depend on temperature, dehydration and damage. Evaporation, perfusion and cooling effects are considered. However, this amount of realism is reflected in computation
times in the range of several hours. For the simulation of MWA the absorption of microwave energy by the tissue has to computed [208].

The first work to investigate the utilization of a thermal necrosis simulation for the purpose of tumor volume coverage optimization for RFA planning is presented by Altrogge et al. [34]. The used simulation model utilizes the finite element method (FEM) and considers the cooling effects of surrounding blood vessels. A multi-scale gradient descent method is used to find the optimal solution resulting in computational times in the range of several hours. Chen et al. [55] also use FEM simulation of the necrosis resulting from RFA for a gradient descent based optimization of the instrument position. Their strategy for the reduction of the computational effort is based on the assumption that over small variations of instrument parameters, the necrosis changes slowly. Hence, during optimization a precomputed temperature field is reused for many cycles and updated after a certain number of iterations. However, the method has only been applied to simple models and the computational times are still in the range of 2 hours.

The computational effort for the prediction of the ablation zone can be reduced by utilizing approximative approaches. A first method that tries to incorporate the heat sink effect of nearby vessels is described by Villard et al. [194]. It deforms an initial ellipsoid based on the proximity to vascular structures. To this end, the vertices of the ellipsoid surface are simply pushed back along their normal vectors in order to maintain a minimum distance to the vessels. Kröger et al. [116] propose the first approximation method that is based on numerical simulation. The main idea of their approach is the subdivision of the problem into a patient independent part and a patient dependent part. The patient independent part comprises the precomputation of the cooling effect for a large number of reference configurations. The patient-specific ablation zone is reconstructed from these stored configurations based on the Euclidean distance of the instrument’s electrode to the vasculature as well as the local vessel radii. Similar to the approach by Villard et al., this method makes strong assumptions about the base shape of the necrosis: it is represented as a union of spheres. To overcome this limitation, Rieder et al. [12] propose a model based on weighted distance fields to approximate the ablation zone under consideration of the heat sink effect. The weighted distance fields are derived from numerical simulations. To consider the cooling of blood vessels, the thermal equilibrium of the vasculature is computed in a preprocessing step. The final ablation zone is the result of a heuristic combination of both fields and can be computed in real time. The result is very similar to the numerical simulation. However, such realistic approximations have not been used for the purpose of optimization yet. Instead, the ablation zone is typically represented by simple spherical or ellipsoidal models based on specifications by the instrument vendors [193, 38].

In addition to target volume coverage, further clinically relevant criteria as described in Section 2.2.2 have to be considered to compute access path proposals of practical relevance. The incorporation of vital structure avoidance is pioneered by Vaillant et al. [188] in the field of computer assisted planning of keyhole neurosurgery, which represents a similar setting. This work and later publications from the field [31, 175]
compute the risk of all possible paths based on the distance to critical structures along the path and either visualize it on the brain surface for manual path selection [188] or use it to automatically determine an optimal path [31, 175]. The first work to combine risk structure avoidance and tumor coverage optimization for the planning of tumor ablation is presented by Villard et al. [193]. Their method uses a downhill simplex optimization scheme with computing times in the range of a minute to obtain an optimal coverage of the target volume by the ablation zone which is approximated by an ellipsoid. Candidate paths that penetrate vital structures are excluded. However, this constraint is too strong and sometimes restricts the optimization process to a zone that is surrounded by anatomical structures. This work also proposes the manual definition of an insertion window to exclude impractical access paths and discusses the avoidance of strongly angulated paths. A subsequent publication by the same authors [192] proposes the usage of projections centered at the tumor to automatically determine regions of interest on the skin under consideration of risk structures. For each insertion zone, the optimization is carried out separately. In a following publication, the same group [39] proposes a precise determination of regions of interest with respect to four strict constraints (penetration depth, liver capsule penetration angle, penetration depth in healthy liver tissue, avoidance of risk structure penetration). The accuracy of the insertion zones as well as the duration of the computation (4-130 s) strongly depend on the resolution of the polygonal representations. In a subsequent publication [38], the authors combine this region determination method with a downhill simplex based optimization method that optimizes a single instrument with respect to three additional soft constraints (optimization of target volume coverage based on ellipsoidal approximation, maximization of distance to risk structures, preference of short paths) inside the precomputed regions. The mean duration for the optimization excluding the region determination is stated to be 30 s. This approach is later adapted for deep brain stimulation [72], which involves very similar objectives, except the need to cover the tumor. Seitel et al. [172] extend the work of that group. Their work concentrates on geometrical criteria regarding the path. Target volume coverage is not considered. The concept of Pareto-optimality is utilized to allow for better exploration of the solution space. However, their approach and its implementation especially with respect to this aspect are described in a very vague manner. Recently, Pareto-based access path optimization was also introduced into the field of neurosurgery planning. The method by Hamze et al. [91] is based on multi-objective evolutionary algorithm which utilizes a Pareto ranking scheme. The same group also investigates the optimization of access paths under consideration of the deformation of the patient anatomy as well as the instrument [92].

4.3 Problem Definition

The goal of the optimization problem discussed here is to identify an optimal access path \( x \) which is defined by a target point \( t \in \mathbb{R}^3 \) in the target volume \( T \) and a
Figure 4.1: The definition of a direction \( d_1 \) illustrated in three-dimensional cartesian space (a). The human figure illustrates the relation of the directions to the patient coordinate system. Figure (b) shows the direction \( d_1 \) in the two-dimensional spherical coordinate system.

direction \( \tilde{d} \), which can be represented as a point in a two-dimensional spherical coordinate system which is oriented such that the zenith direction is aligned with the positive \( z \)-axis of the patient coordinate system (Figure 4.1 (a)). Hence, \( x \) is given by:

\[
x = (t, \tilde{d}) \quad t \in \mathbb{R}^3, \quad \tilde{d} \in S^2.
\]

For convenience, the direction \( \tilde{d} \) is represented by a vector \( d \in \mathbb{R}^2 \) consisting of the longitude (or azimuth) angle \( \theta \) and the latitude (or elevation) angle \( \phi \):

\[
\tilde{d} \simeq d = \begin{bmatrix} \theta \\ \phi \end{bmatrix} \quad \theta \in [0, 2\pi), \quad \phi \in [\phi_{\text{min}}, \phi_{\text{max}}].
\]

The range of \( \phi \) is restricted to \( [\phi_{\text{min}}, \phi_{\text{max}}] \). Hence, trajectories with an angle to the transverse plane above a certain threshold are excluded. These trajectories are of no practical value for interventions in the liver since they would traverse through the head or the pelvic area. To this end, thresholds of \(-\frac{1}{3}\pi\) and \(\frac{1}{3}\pi\) are used for \( \phi_{\text{min}} \) and \( \phi_{\text{max}} \), respectively. The longitude is measured as the angle to the positive \( y \)-axis of the patient coordinate system.

Alternatively, the direction \( d = [\theta, \phi]^T \in \mathbb{R}^2 \) can be represented by a three-dimensional vector in \( \mathbb{R}^3 \) as follows:

\[
d_R = [-\sin(\theta), \cos(\theta), \tan(\phi)]^T \in \mathbb{R}^3
\]

Hence, a path \( x \) might be defined by \( (t, d) \) or \( (t, d_R) \). Since both representations are equivalent, \( \theta \) and \( \phi \) can also be calculated based on \( d_R = [d_{Rx}, d_{Ry}, d_{Rz}]^T \):

\[
\theta = \arctan\left(\frac{d_{Rx}}{d_{Ry}}\right)
\]
4.3. Problem Definition

\[ \phi = \arccos \frac{d_{Rz}}{\sqrt{d_{Rx}^2 + d_{Ry}^2 + d_{Rz}^2}} \]  

(4.5)

Note, that the inverse tangent as used in Equation 4.4 must be suitably defined to be valid for all directions in the transverse plane. Most programming languages offer a function \texttt{atan2}(x,y) for that purpose.

Figure 4.2 illustrates the geometry of a trajectory \( x \) and the surrounding anatomy. Each trajectory \( x \) penetrates the boundaries of the target volume, the liver, and the skin at the points \( p_t(x) \), \( p_l(x) \), and \( p_s(x) \), respectively. The orientation of the liver surface at \( p_l(x) \) is given by \( n_l(x) \). \( S \) denotes the risk structures to be considered:

\[ S = \bigcup_{i=1}^{n} S_i \]  

(4.6)

where \( n \) is the number of risk structures. The criteria outlined in Section 4.1 can be formulated as follows:

- **\( f_1(x) \)** - the penetration depth of \( x \) is given by:
  \[ f_1(x) = f_1(t, d) = \|p_s(t, d) - t\| \]  
  (4.7)

  This measure describes the distance between target and skin entry point. Depending on the exact implementation, the distance between target point and instrument tip might be added.

- **\( f_2(x) \)** - the distance of \( x \) to the closest risk structure is given by:
  \[ f_2(x) = f_2(t, d) = \min_{0 \leq l \leq l_{max}} \|s - (t + ld_R)\|, \]  
  (4.8)

  where \( l_{max} \) is a constant, for example the length of the instrument.

- **\( f_3(x) \)** - the angle between \( x \) and the transverse plane is given by:
  \[ f_3(x) = f_3(t, [\theta, \phi]^T) = \phi \]  
  (4.9)

  If the direction of the path is given as a three-dimensional vector \( d_R \), \( \phi \) is calculated using Equation 4.5.

Figure 4.2: Illustration of the geometry of a path \( x = (t, d) \) and the surrounding anatomy.
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- $f_4(x)$ - the angle between $x$ and the y-axis in the transverse plane is given by:
  \[ f_4(x) = f_4(t, [\theta, \phi]^T) = \theta \] (4.10)
  If the direction of the path is given as a three-dimensional vector $d_R$, $\theta$ is calculated using Equation 4.4.

- $f_5(x)$ - a measure describing the probable therapy outcome for a path $x$ can be defined in various ways, for example by computing the overlap of the probable ablation zone $C(x)$ with the target volume $T$:
  \[ f_5(x) = \frac{|T \setminus C(x)|}{|T|} \] (4.11)

- $f_6(x)$: the length of the segment of $x$ that penetrates healthy liver tissue (transhepatic route) is given by:
  \[ f_6(x) = \|p_l(x) - p_t(x)\| \] (4.12)

- $f_7(x)$: The angle between $x$ and the liver capsule is given by:
  \[ f_7(x) = f_7(t, d_R) = \arccos \frac{d_R \cdot n_l(t, d_R)}{|d_R|} \] (4.13)

The formulation of the respective optimization problem may vary depending on the used algorithm and approach (see Section 2.5). In the following, two methods for the MOO using the objectives defined above including the implemented solution are presented.

4.4 Fast Automatic Optimization of Trajectories

The first approach for MOO of access paths for the planning of percutaneous intervention in the liver is based on the assumption that the target point $t$ is clearly defined, for example by the center of the target volume. As described in Section 3.4.12, this assumption is valid if the ablation zone is large enough to cover the target volume and heat sink effects are neglected. The method presented here exploits this assumption in order to allow for almost instantaneous computation of clinically relevant access path proposals. In Section 4.5 and Section 4.7 extensions of the introduced methodology are presented which allow to optimize the target point as well under consideration of a more plausible model of the ablation zone including heat sink effects. However, in both cases longer computation times are required.

In this section, the identification of a trajectory $x$ for a fixed target point $t$ is considered only. Hence, only the direction $d$ has to be optimized. For that, the method presented here introduces an efficient brute force computation of the objectives $f_1 \ldots f_7$ and combines it with an a priori articulation of preferences based on utility functions.
The proposed method requires the input of segmentation masks representing the anatomical structures that have to be considered. In addition to a risk structure mask, which labels all voxels that belong to a risk structure, a mask discriminating background voxels from those that belong to the patient (patient mask), and a liver mask are required.

In the following, the method is described in detail. Afterwards, results, including two clinical studies, are presented and discussed.

### 4.4.1 Brute Force Objective Computation

The assumption of a fixed target point reduces the optimization problem to the search for an optimal trajectory to the fixed target point \( t \). Hence, all possible trajectories correspond to points in a spherical coordinate system (see Figure 4.1 (b)). The feasible set is discretized by regular subdivision of this two-dimensional, rectangular coordinate system in both directions \( \theta \) and \( \phi \). Due to the regular sampling in the latitude and longitude dimensions, the objective values can be stored as two-dimensional images, which will be referred to as objective maps for the remainder of this work. Depending on the chosen resolution, the number of trajectories, and therefore the number of objective computations, is very large. An angular resolution of one degree would result in \( 360 \times 120 = 43200 \) trajectories. Hence, to allow for practicable computation times, the following two approaches are used:

**Computation based on two-dimensional trajectory coordinates:** For some of the objectives the function values can be derived directly from the direction of the trajectory. Therefore, values for each trajectory are assigned based on the longitude and latitude of the trajectory. This is the case for the objectives \( f_3 \ldots f_5 \).

**Computation based on analysis of three-dimensional patient anatomy:** Objectives, that require an evaluation of the patient anatomy for each trajectory \( x \) are computed utilizing the GPU. This is the case for \( f_1, f_2, f_6 \) and \( f_7 \). For each of these objectives an equirectangular projection with the center at the target point is computed. This variant of cylindrical projection maps longitude and latitude to equally spaced lines in a rectangular coordinate system. Hence, it is also called latitude-longitude projection [88]. The projection is carried out on the GPU in two steps:

1. **Generation of a cube map:** As described in Section 2.4, cube maps can be used to capture all directions to one point. The projection center of the cube map is the target point. DVR (see Section 2.4) is used to render the patient information into the map. The rendering is adapted in order to output the desired objective values for each direction.

2. **Mapping from the cube map to the spherical coordinate system:** For each pixel \( p \) in the longitude / latitude coordinate system a cube map lookup vector
\( v_p \) is computed using Equation 4.3. The value of the pixel \( p \) is determined by fetching a value from the cube map using \( v_p \).

The result of the projection is transferred from the GPU to the central processing unit (CPU) in order to allow for further processing. In the following, the map generation for the seven objectives utilizing the two approaches is described.

**Penetration depth \((f_1)\):** Inspired by the depth buffer (see Section 2.4.3), the depth of the patient skin is rendered into the objective map during the GPU based projection based on the patient mask. To achieve this, the vertex and fragment shaders of the DVR are modified to output the distance between skin entry point and the target point. The resulting map stores this distance in millimeters for each trajectory (Figure 4.3 (a)).

**Distance to the closest risk structures \((f_2)\):** This objective is computed using the GPU based projection approach. The resulting objective map stores for each trajectory the distance of that trajectory to the closest risk structure. To achieve this, a Euclidean DTF \([49, \text{ p.270}]\) based on the risk structure mask is computed first. Since this computation is completely independent of the target point, it has to be carried out only once for each patient. Hence, it can be implemented as a preprocessing step. To generate the objective map itself, the equirectangular projection of the minimum intensity projection of the distance map is computed using DVR (Figure 4.3 (b)) during the GPU based projection (see Section 2.4.1). The resulting Euclidean distances are given in millimeters.

**Angulation to transverse plane \((f_3)\):** The angulation to the transverse plane corresponds to the latitudinal angle. Therefore, a map is generated based on two-dimensional trajectory coordinates. It stores the respective latitude value in each pixel. All pixels of one row have the same value, because the respective trajectories have the same latitude. Because angulation in the superior direction (pointing upwards, more towards the head of the patient) might be treated differently than angulation in the inferior direction (pointing downwards, more towards the feet of

Figure 4.3: Examples of objective maps generated using GPU based projection and volume rendering. \((a)\): The penetration depth objective map. \((b)\): The risk structure distance objective map.
the patient), signed values are stored. Therefore, the values range from $-\frac{1}{3}\pi$ to $\frac{1}{3}\pi$ where negative values represent inferior angulations.

**Angle to the y-axis in the transverse plane ($f_4$):** This map stores the longitudinal angle of each trajectory. Similar to $f_3$, the map values are derived directly based on the coordinate of the pixels, because the longitude of a trajectory corresponds to the column of the respective pixel. The values range from 0 to $2\pi$.

**Tumor volume coverage ($f_5$):** This objective map is generated generically. A simple geometric heuristic is used to determine for all trajectories the probability for a good target volume coverage. It is based on the following observations:

- Those trajectories that are close to the main axis of the target volume facilitate a good coverage of the target volume for instruments with longish ablation areas such as needle-shaped applicators (Figure 4.4 (b)).

- For instruments with an ablation area of flat shape (e.g. umbrella-shaped electrodes), those trajectories are preferred too, because they permit ablation of the whole target volume using the pull-back technique (see Section 2.2.2, Figure 4.4 (c)).

Therefore, the deviation to the main tumor axis is encoded for each trajectory in the map. Volume rendering is not required for this. Instead, simple image processing is utilized to compute the deviation for each trajectory. In the first step, the spherical coordinates of the two trajectories that are parallel to the tumor axis are computed (Figure 4.4 (a)). These coordinates are used as centers of two radial distance fields. Both fields are combined using a minimum operator and stored

Figure 4.4: The computation of the tumor coverage objective is based on the analysis of the main tumor axis. (a): There are two trajectories (red arrows) that lie parallel to the main tumor axis (dashed line). Coverage of the tumor can be achieved with one longish (b) or multiple ablations on the trajectory (c). (d): The resulting objective map for the computed main tumor axis. The two trajectories that are parallel to the axis (red points) are outside the covered range of $\phi \in (-\frac{\pi}{3}, \frac{\pi}{3})$ in this example.
into the map (Figure 4.4 (d)). A special border handling ensures that parts of the distance fields beyond 360° or 0° of longitude are shifted adequately. Because the image domain is a 2D spherical coordinate system, the resulting pixel values of the map are the deviations from the main tumor axis in degrees. This objective is only used for tumors of a considerably elongated shape.

**Length of healthy liver tissue segment** \( (f_6) \): For projection of the length of the segment of the path that penetrates healthy liver tissue a depth map is used in a similar way as for the computation of \( f_1 \). However, the result is composed of two passes: the depth of the volume rendering of the liver mask and the tumor mask are computed separately and subtracted from each other afterwards.

**Liver capsule penetration angle** \( (f_7) \): The shaders of the DVR are modified to render the angle between the normal vector of the liver surface and the direction of the trajectory into the map during GPU based projection. The normal vector is computed based on a smoothed liver mask, because gradient estimation for binary images as well as strong anisotropic voxels can be problematic. Hence, the mask image is converted to a floating point image, resampled to an isotropic voxel size and finally smoothed using Gaussian smoothing [85]. This derivation of the smoothed liver mask can be integrated into a preprocessing step.

In the following, the mapping of the values of these seven objectives to a common range, and the combination into one scalar value to allow for optima extraction, are described.

### 4.4.2 Scalarization

As described in Section 2.5, a common approach to MOO is to scalarize the vector of objectives in order to be able to find an optimum. To this end, an a priori method from the field of *utility theory* is applied. A *utility function* \( U : \mathbb{R}^k \to \mathbb{R} \) maps the value \( f(x) \) of the design (i.e. \( x \)) to a scalar:

\[
U(f(x)) = U(f_1(x), f_2(x), ..., f_k(x)) \quad (4.14)
\]

However, typically this function is composed of multiple individual utility functions \( U_i(f_i(x)) : \mathbb{R} \to \mathbb{R} \) which express the value of the design as a function of the attribute expressed by \( f_i(x) \). The seven objective maps \( f_i \) contain values for each possible trajectory in adequate units and value ranges for the respective objective. However, these values do not necessarily correlate directly to wanted or unwanted values. Therefore, the \( U_i \) are designed such, that these values are mapped to \([0, 1]\), whereas low values indicate unwanted and high values favored characteristics. Weighted utility functions \( U_i^{-} \) are constructed using weighting factors \( w_i \in \mathbb{R}, \ 0 \leq w_i \leq 1 \) in order to adapt the importance of single objectives without having to manipulate the curve directly:

\[
U_i^{-}(f_i(x)) = U_i(f_i(x))w_i + 1 - w_i \quad (4.15)
\]
4.4. Fast Automatic Optimization of Trajectories

The two rightmost terms are added to scale the function while keeping its maximum. A combination of the individual utility functions can be achieved in various ways, for example by using a sum or product [36]. The weighted sum is very common. However, it allows for compensation of unwanted objective values by favored values resulting from other objectives. Hence, for the problem at hand, the product is used:

\[
U(f(x)) = \prod_{i=1}^{7} U_i(f_i(x)) \\
= \prod_{i=1}^{7} [U_i(f_i(x)) w_i + 1 - w_i]
\] (4.16)

Since the utility function expresses the merit of a certain solution for the therapy, high values are preferred. Hence, \(U(f(x))\) is maximized [36]. As a result, the introduced weighting factors \(w_i\) have the following consequences:

- If for one objective \(f_i\) the weight \(w_i\) is 1 and \(U_i(f_i(x))\) is 0 for a trajectory \(x\), then \(U_i(f_i(x))\) is 0. Since \(U(f(x))\) is a product, this results in an overall utility value of 0. Practically, \(x\) would be excluded as a possible solution. Hence, objective \(f_i\) has a high importance and constraints the feasible design space additionally.

- If for one objective \(f_i\) the weight \(w_i\) is 0, then the value of \(f_i(x)\) is of no consequence and \(U_i(f_i(x))\) is 1 for all trajectories. Hence, objective \(f_i\) has no influence.

Since the objective maps are images, the application of the individual utility functions as well as the product are implemented as image processing operations. Hence, the resulting \(U(f(x))\) is also an image.

Figure 4.5 illustrates the utility function for the angulation objective map \(f_3\). The original objective map contains values from \(-\frac{\pi}{3}\) to \(\frac{\pi}{3}\). The function \(U_3\) assigns values between 0 and 1. The peak of the function is at 0° in order to favor trajectories that are in plane. Negative values of \(f_3\) correspond to inferior angulations and are better suited than superior angulations. Hence, the function is not symmetrical.

![Figure 4.5: Utility function for the angulation objective. (a): The original map \(f_3\) contains values from \(-\frac{\pi}{3}\) (bottom) to \(\frac{\pi}{3}\) (top). (b): The corresponding utility function \(U_3\) assigns values between 0 and 1. (c): It favors trajectories close to the transverse plane and penalizes superior angulations stronger than inferior angulations.](image-url)
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Figure 4.6: Illustration of the utility function for $f_4$ (angulation in the transverse plane): Different restrictions apply to the posterior right (a), the anterior right (b) and posterior left (c) part of the liver.

The implementation of the utility function $U_4$ for the objective $f_4$, which expresses the angle of the trajectories to the y-axis in the transverse plane, differs from the other utility functions, since the rating of certain directions depends on the position of target inside the liver. To this end, variants of $U_4$ are defined for the four quadrants in the transverse plane. The variant for the quadrant in which the target point is located is chosen (see Figure 4.6).

4.4.3 Determination of Optimal Paths

The result of the scalarization is an image containing $U(f(x))$ for all trajectories $x$. Hence, the maximum of $U(f(x))$ is can be determined using simple image processing operations. First, a Gaussian smoothing [85] is applied. Thereafter, all local maxima in a 8-neighborhood are extracted and sorted based on their value in $U(f(x))$. The maximum with the highest value is initially selected, but any other maximum from a sorted list of restricted size may be chosen by the user. The longitude and latitude values of the selected maximum are used to compute the direction of the corresponding access path. The skin entry point is computed based on the given target point, the spherical coordinates of path direction and the corresponding penetration depth, which can be looked up in the objective map representing the penetration depth. The resulting path can be visualized by drawing a line or a virtual applicator model. An overview of the complete algorithm is given in Figure 4.7.

4.4.4 Results

The proposed method is implemented with focus on CT-guided RFA and biopsy using the rapid prototyping platform MeVisLab [164, 139] and integrated into a clinical software demonstrator.

Utility function determination: During several workshops with radiological experts, the objectives and the respective utility functions and weighting factors are
4.4. Fast Automatic Optimization of Trajectories

Figure 4.7: Overview of the proposed projection based algorithm: Objective maps $f_i(x)$ are generated based on the input masks (blue) and preprocessing results (green). To each objective a merit is assigned by an according utility function $U_i$. The final scalar function $U(f(x))$ is yielded by a weighted product. From the resulting image, the maxima are extracted. For each maximum the skin entry point is computed. The user can iterate through the resulting list of optimal access paths.
discussed. Based on this expert knowledge, the algorithm and its parameters are adjusted in an iterative process. To verify the suitability of the chosen utility functions and weighting factors, an additional investigation of the path parameters of actually placed applicators is conducted. This analysis covers 37 different access paths defined by three radiologists on 19 RFA CT planning scans. After manually deriving the path parameters for each access path, the average value as well as the upper and lower bound for each parameter are calculated. The previously defined utility functions are compared with these values and adjusted when necessary. The exact results of this process are confidential. The project sponsor did not agree to the publication of these informations. Hence, they are not contained in this thesis.

Segmentation and preprocessing: The software demonstrator integrates several methods for the segmentation of structures that are needed as input for the proposed trajectory optimization method. An automatic preprocessing step segments the skin, ribs, lung and liver after the planning CT data has been imported. For the segmentation of the liver, a proprietary algorithm (Siemens Corporate Research, Princeton, USA, [114]) is used. Segmentation of the skin, ribs, lung, and air-filled parts of the bowel and the stomach is based on thresholding. Since the pulmonary recessus is not entirely visible in the CT data, it is approximated based on the lung mask using model assumptions. The according algorithm is treated as confidential due to restrictions imposed by the project sponsor. The segmentation of the tumor is carried out in a semiautomatic fashion by the user. To trigger the segmentation, a line is drawn along the largest diameter of the lesion. The method of Moltz et al. [140] is used for the segmentation of the lesion based on this input. The preprocessing step, which computes the Euclidean DTF based on the risk structures and the resampled smoothed liver mask, is carried out automatically after the automatic segmentation step. Its mean duration is 4 seconds for abdominal scans of moderate size (512 x 512 voxels per slice, 44 - 203 slices) on a standard laptop computer (Intel Core2 Duo T9500 @ 2.6 GHz, 3 GB Ram, NVIDIA GeForce 8600M GT, Windows Vista 32Bit).

Access path definition: Once the physician has segmented a tumor, he can trigger the automatic computation of access path proposals by pressing a button. The computation is performed in less than one second. The best proposal is automatically visualized by means of a virtual instrument. The physician can move the tip of the instrument, if necessary, and the proposals are updated as soon as the interaction is finished. Again, this is executed in less than one second. After each update, the physician can step through the list of proposals.

4.4.5 Evaluation

Two studies are carried out by clinical partners to assess the value of the presented method. The first study focuses on RFA in the liver, the second study on liver biopsy.
Although the algorithm parameters are determined with focus on RFA, they are also appropriate for biopsy planning, because except for tumor coverage, the radiologists consider the same criteria during planning.

**Study 1** The first retrospective study includes CT data of 26 patients (12 men and 14 women; age range 39-86 years; mean age 63 years) which were treated by CT-guided RFA of malignant liver tumors (18 metastases and 10 HCC). The CT data is obtained from three different university centers of radiology and contained pre- and intraprocedural images. The segmentation and planning is carried out based on the preprocedural CT datasets. All necessary steps are integrated into a study software that allows for efficient implementation of the study.

Before the computation of the automatic access paths, a standard of reference is established by means of several experts’ proposals. For that, optimal access paths are manually determined by three radiologists with a six year experience in interventional radiology for each of the datasets (see Figure 4.8) during a workshop. After this workshop, one of the three radiologists retrospectively evaluates the proposed method in order to determine the quality of its results. For this, he first manually defines the target position inside the tumor. Three path proposals are then generated by the method. The radiologist rates the results in an absolute and a relative fashion. The absolute evaluation involves a rating of the three paths per case using a six-point Likert scale (one=very good, two=good, three=satisfactory, four=sufficient, five=insufficient, six=fail), taking into account instrument angulation, distance to risk structures, and penetration depths regarding both skin and liver. The mean

Figure 4.8: Definition of the three expert path proposals *(red, blue and green lines)* in the study software before the automatic path proposal generation.
score of the marks assigned to the automatically computed path proposals is 1.77. The best mark is one and the poorest mark is three. Hence, all proposals are feasible and rated at least sufficient. The mean score for the first, second and third computed path proposal is 1.65, 1.96, and 1.69, respectively.

For the relative evaluation, the first of the three automatic path proposals is compared with the three anonymous expert proposals using a four-point Likert scale:

- 1: feasible and better than experts’ proposal
- 2: feasible and equally good
- 3: feasible, but less suitable than experts’ proposal
- 4: not feasible

This leads to three relative scores per path. An overall score per path is derived by using the worst of the three scores. Compared to the three “gold standard” path proposals, in five of the cases (19%) the best automatic proposal is evaluated as being better. In 19 cases (73%) they are evaluated as equal and in two cases (8%) they are, feasible, but not as good as the experts’ proposal. Figure 4.9 shows an example of a case that required a double oblique access. The automatic proposal closely matches the path that was used in the actual intervention.

Study 2 This retrospective study includes 33 patients (21 men and 12 women; age range 19–82; mean age 60 years) referred for CT-fluoroscopy-guided biopsy of focal liver lesions. Two patients are excluded from the study, because of failure of the automatic segmentation process caused by significantly altered shape of the liver due to prior liver resection. Among the remaining 31 patients, there are 33 focal liver lesions (21 in the right, and 12 in the left liver lobe).

The segmentation and planning is carried out based on the preprocedural CT datasets. In addition to the common segmentation pipeline, the epigastric vessels are segmented. For that, several markers are placed along the course of the vessel. The segmentation is performed using a combination of a vesselness filter and a fast marching algorithm, similar to the methods proposed by Cai et al. [51] and Friman et al. [79]. After segmentation, the proposed method is used to generate path proposals for a user-defined target point in the tumor. The three best proposals are presented in a sorted order, starting with the best proposal.

Two experienced radiologists (five and ten years of experience) assess the results based on a manually defined optimal needle path by consensus decision, which serves as the standard of reference (experts’ proposal). Each of the three automatic path proposals is rated as:

- comparable to experts’ proposal
- feasible, but less suitable than experts’ proposal
- not feasible
4.4. Fast Automatic Optimization of Trajectories

Figure 4.9: Example of a path proposal (red) for a lesion beneath the liver dome (visible in Figure (a), orange, black outline): the trajectory is double oblique, the skin entry point is in another slice (b) than the target point (a). Segmentation masks for high density (green) and low density (yellow outline) risk structures are visualized as overlays. Figure (c) shows one slice of the intraprocedural scan with the actually placed instrument for comparison.
In all 33 cases, the first automatic path proposal is at least rated feasible. In 25 of 33 cases (76%), it is rated comparable to the experts’ proposal. Figure 4.10 shows an automatic path proposal, that is comparable to the experts’ proposal. To avoid crossing the recessus of the lung, an angulation of the trajectory in z-axis is necessary. Figure 4.11 shows an automatic path proposal, considered not feasible, that closely passes several risk structures. In 30 of 33 cases (91%), one of the three automatic path proposals is rated comparable to the experts’ proposal. Hence, in several cases, the second or third proposal is better then the first. Seven of the overall 99 proposals (7%) are rated not feasible, one of them being a second proposal, and six of them being the third proposal (see Table 4.1).

In this study, also the quality of the complete segmentation process is evaluated in a consensus reading by the two radiologists. The results are rated using a six-point Likert scale (one=very good, two=good, three=satisfactory, four=sufficient, five=poor, six=deficient). In 26 cases (79%), all automatic and semiautomatic segmentation results are at least good (average grade for segmentation is excellent for skin contour, very good for lungs, intestine, bone, epigastric vessels, and good for liver and focal liver lesions). In seven of 33 cases (21%), one or more automatic or semiautomatic segmentation results are rated insufficient. An example of a good automatic segmentation process with good delineation of the liver and the dorsal recessus, as well as an example of a deficient automatic segmentation process of the liver contour, are shown in Figure 4.12. Semiautomatic segmentation of the epigastric vessels (if relevant with respect to the trajectory) as well as the focal liver masses (selected for CT-guided biopsy) is successfully performed in all 33 CT datasets. Due to the thresholding used for the automatic bone segmentation algorithm, other structures that are characterized by a high density/attenuation including the aorta, kidneys, and pancreas in the arterial phase may be included in the segmentation. This is a technical limitation of the automatic bone segmentation algorithm (compare Figure 4.10, (a, b) and Figure 4.12, (b)). Nevertheless, in this study, these structures are not within the proximity of a potential access trajectory and have no influence on the path proposals.

<table>
<thead>
<tr>
<th></th>
<th>Proposal 1</th>
<th>Proposal 2</th>
<th>Proposal 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Comparable to experts’ proposal</td>
<td>25 (76%)</td>
<td>15 (45%)</td>
<td>13 (39%)</td>
</tr>
<tr>
<td>Feasible, but less suitable than experts’ proposal</td>
<td>8 (24%)</td>
<td>17 (52%)</td>
<td>14 (42%)</td>
</tr>
<tr>
<td>Not feasible</td>
<td>0 (0%)</td>
<td>1 (3%)</td>
<td>6 (18%)</td>
</tr>
</tbody>
</table>

Table 4.1: Results of study 2: evaluation of the automatic path proposals.
4.4. Fast Automatic Optimization of Trajectories

Figure 4.10: Results for exemplary case with successful automatic path proposal computation: the skin (orange), hyperdense structures (bones, calcifications, enhanced vessels, and renal cortex, yellow), air-filled organs including the recessus of the lung (blue), and the liver (green) are automatically segmented, whereas for focal liver lesion (white) and epigastric arteries (not shown), semiautomatic segmentation is used. The best automatic path proposal (yellow), is comparable to the experts’ proposal (green). For not crossing the recessus of the lung, an angulation of the trajectory in z-axis was necessary. Both paths also resemble more or less the path that was chosen for the biopsy (c, d).
Figure 4.11: Example of a case with infeasible proposals: the proposed path (yellow) violates intrahepatic vascular structures (c). This was caused by the fact that these structures were not segmented.

4.4.6 Discussion

In contrast to many other methods in the field of automatic access path determination, the presented approach does not focus on the volume coverage problem only. Instead multiple criteria are considered to compute appropriate access paths for a given target point. In that regard, it is mainly comparable to the works of Baegert et al. [38] and Seitel et al. [172]. However, the presented method is faster due to the application of techniques from the fields of computer graphics and image processing as well as the utilization of the graphics hardware. An additional difference of the presented method to most other works from this field is that an additional polygonization step is not required since all objectives are computed based on segmentation masks directly. This does not only remove the associated computation time, errors and algorithm parameters, it also facilitates integration into existing software.

In the two retrospective clinical studies, the clinical value of the proposed methods
Figure 4.12: A good segmentation result of the liver contour even including the small indentation at the medial inferior liver margin (a). In contrast, (b) shows a deficient segmentation result of the liver contour leaving out a large area of segment 6 at the dorsal liver margin. Though, as the delineated tumor is located distant from the deficient segmentation area, there is no negative effect on the 3 calculated automatic path proposals. Automatic path proposal (continuous yellow line) and experts’ path proposal (continuous green line) are in plane.

has been evaluated. The studies have several limitations. First of all, the number of patients in both studies was limited (26 and 33 cases). Second, the clinicians that rated the results were also involved in the definition of the standard of reference. Nonetheless, the results are promising. The proposed method enabled a rapid automatic path proposal that was comparable to the experts’ proposal in the majority of cases. Infeasible proposals mainly resulted from missing or inaccurate representation of risk structures. Nevertheless, even for suboptimal segmentation results the proposed paths were feasible in many cases because local segmentation defects only have a local influence. However, structures that are not present in the risk structure mask at all, may be penetrated by trajectory proposals (see Figure 4.11). Hence, an application of the proposed method should be based on complete and accurate segmentation masks, if possible.

With regard to the classification of optimization methods given in Section 2.5, the presented method uses priori articulation of preference information by means of utility functions and weighting factors. The application of these methods to the intervention planning problem has been recently criticized. It is seen as a disadvantage that the weights and utility functions have to be set by the user. Seitel et al. [172] argue that the user would have to do this every time. However, the incorporation of utility functions and weight factors in the presented work aims at modeling the preferences and knowledge of the experts involved in the project. The method is mainly used without modifying these parameters, for example by less experienced
radiologists. Alternatives should mainly be selected from the sorted list of results, which allows for posteriori articulation of preference information. It should be noted though, that the inferior solutions from the sorted list are not necessarily Pareto optimal. This interaction was still found useful because it allows for the selection of local optima that would most probably have been global optima for slightly modified weight factors. Due to the brute force computation of all objectives before the actual scalarization, the method also allows for interactive modification of the utility functions and weighting factors. Hence, the prior articulation of preference information could also be used in a way that is similar to a posteriori articulation of preference information. However, it is known that the modification of weight factors of scalarization methods might not always lead to a predictable behavior. Small changes of weights might result in strong changes of the optimum. Hence, other approaches to interaction approaches for the interactive exploration of the Pareto set should be investigated.

The required definition of the target point poses no limitation of the practical application of the method for biopsy or radiofrequency ablation of small lesions. In most cases the lesion center can be used as target point and then refined manually by the radiologist. The evaluation studies show that the method is suited to support the planning of such interventions in the liver and that its value is not hampered by the assumption of a user defined target point. It still represents an improvement over manual planning since it augments the search for a suited access. For a further improvement of the planning a realistic estimation of the ablation zone should be integrated. The utilized heuristic does not incorporate a realistic estimation of the ablation zone and the resulting coverage of the target volume. It only tries to identify one main direction which might allow for satisfying overlap based on a purely geometric analysis.

4.5 Pareto-Navigation for Optimization of Access Paths

In the field of intensity-modulated radiotherapy planning, multi-criteria optimization based on Pareto surface navigation is an established method [118]. This concept encompasses methods that allow for an interactive exploration of the Pareto front. Hence, patient individual decisions can be made based on interactive trade-off definition. The final decision for the therapy therefore lies in the hands of the physician. It has been shown that this approach allows to reduce the average duration of radiotherapy planning from 135 to 12 min and it seems that the generated plans are superior to manual planning results [60]. As pointed out by Jolesz, ”Planning for thermal ablation and radiation therapy has substantial similarities” [101]. Hence, a translation of that concept to the planning of tissue ablation procedures seems natural. However, no comparable systems have been devised so far. Furthermore, none of the developed methods for multi-criteria planning of tissue ablation procedures
incorporate a realistic representation of the thermal necrosis.

In this section, a new method for the efficient patient-specific determination of suited access paths for hepatic RFA based on segmentation masks is proposed. Similar to the work by Seitel et al. [172], the concept of *Pareto optimality* is utilized as the guiding principle for the selection of access paths considering multiple clinically relevant criteria. However, the proposed method allows for intuitive exploration of the set of Pareto-efficient solutions inspired by works in the field of intensity-modulated radiotherapy planning [118].

The method proposed here is partially based on the work presented in Section 4.4. However, in contrast to that work, the novel approach described here optimizes both, the target point and the direction of the access path. Furthermore, a realistic representation of the ablation zone including cooling effects is used during optimization.

### 4.5.1 Method Overview

The proposed approach considers a subset of the objectives described in Section 4.3: only the objectives $f_1 \ldots f_5$ are implemented. The objectives $f_6$ and $f_7$ are excluded since an automatic segmentation of the liver is not available in many cases. Hence, the method requires segmentation masks of the tumor, hepatic vessels, the patient, the lung and the bones as input.

**Black box algorithms:** Several important components of the overall method are not part of this thesis and are treated as black box algorithms. The integration of these components influences the structure of the overall method. The objective $f_5$, which evaluates the probable therapy outcome, is computed either based on a mathematical simulation of the heat distribution [117] or using the

![Diagram](image.png)

**Figure 4.13:** Overview of the components of the algorithm: The method requires segmentation masks of the considered anatomical structures as input. The black components represent black box algorithms used by the overall method.
ablation zone approximation method of Rieder et al. [12]. Both methods consider the cooling effects of surrounding vascular structures and therefore require a mask representing the vessels as input.

The numerical optimization uses the adapted hyperboxing method by Teichert [184] to calculate a representative set of Pareto efficient solutions, which are interpolated using the interactive method by Küfer et al. [118] for a posteriori articulation to yield the final solution.

The numerical optimization has the following requirements:

- **Starting points:** The optimization algorithm needs a set of starting points that approximate the Pareto front.

- **Objectives:** The optimization requests objective values for candidate access paths. The optimizer follows a minimization approach and the objectives have to be normalized accordingly to the range \([-1, 0]\).

- **Rib cage polytopes:** In order to solve the optimization problems during the Pareto front approximation, the feasible set needs to be restricted in order to prevent the punctuation of bones. For that, the optimization requires an explicit representation of the bones in the form of a set of convex polytopes as input. Other risk structures, such as the lung or vessels are considered during optimization by the objective \(f_2\) (distance to closest risk structure).

Figure 4.13 illustrates the relations between the various components of the proposed method. Both, the starting point creation (see Section 4.5.3) as well as the objective function computation (Section 4.5.4) during the optimization require the computation of distances to risk structures such as lung, bones, cartilage and vessels. Hence, the first step of the overall method is the computation of a Euclidean DTF [49, p.270] based on a mask that contains all relevant risk structures.

Besides the integration of all parts into a method for optimization of access path, this work concentrates on the computation of rib cage polytopes, starting points, and objective function values, which will be explained in more detail in the following.

### 4.5.2 Rib Cage Polytope Computation

In order to solve the optimization problems during the Pareto front approximation, the requirement that the path must not punctuate a bone structure needs to be explicitly formulated in mathematical terms. To obtain this explicit formulation, the bones are approximated by a set of convex polytopes. Hence, the bones have to be subdivided into convex parts. To this end, several image processing steps are used. The resampled bone mask is separated into spine, sternum, and the ribs using morphological operations:

1. First, the spine is detected. A closing operation with a large kernel is used to close gaps resulting from the intervertebral discs and vertebral canal (see Figure 4.14 (b)). Next, an opening removes all thin objects and only the core
of the spine remains (Figure 4.14 (c)). The result is dilated (Figure 4.14 (d)) and used to mask the original mask, which yields the spine (Figure 4.14 (e)).

(2) Next, the sternum is detected based on the resampled bone mask from which the spine is removed Figure 4.14 (f)). Since the sternum resembles a flat object with relatively large dimensions in the x- and z-dimension, it is detected using an opening with an anisotropic kernel Figure 4.14 (g)).

(3) The ribs result from the subtraction of the spine and the sternum from the overall bone mask. To allow for a better separation of the ribs, and to reduce the number of small objects, the sternum is dilated before being subtracted.

The dilated sternum resembles a compact object. Hence, it is represented by one polytope only. Similarly, the spine is represented by one large polytope because it is only marginally bent. Since the gaps resulting from the intervertebral discs and the vertebral canal are closed, and the ribs have been removed close to the vertebrae, the spine approximates a cylindrical object (see Figure 4.14 (e)). For both, the dilated sternum mask and the spine mask, a polytope is created. For that, the convex hull of the mask is computed and converted into polygonal representation, which is smoothed using a volume-preserving Laplacian smoothing operator [154, p.177].

![Figure 4.14: Subdivision of the bone mask](image)
(a) (b) (c) (d) (e) (f) (g) (h)

Figure 4.14: Subdivision of the bone mask (a): Morphologic operations are used to detect the spine (b-e) and subtract it from the original mask (f). Subsequently, the sternum is detected using morphologic operations (g). The ribs (blue) can be separated after the spine and a dilated version of the sternum are removed (h).
Next, the number of polygons is reduced with an algorithm that collapse edges based on quadric error metric [81]. The result is approximately convex. However, some points might be inside the convex hull. These are removed by the numerical optimization component automatically before the polytope is used.

The ribs cannot be represented by one convex polytope. Hence, they are further subdivided. First, individual ribs are extracted by applying a region labeling algorithm [49, p.6]. Next, for each of the individual ribs the centerline is extracted using the approach of Selle et al. [173]. The rib is subdivided by several planes that are perpendicular to the centerline. The distance between the planes is set to 15 mm. The fragments are dilated by one voxel and masked by the original rib mask in order to guarantee an overlap of the fragments. Subsequently, polytopes are generated for each of these fragments in the same way as described before.

### 4.5.3 Starting Point Generation

The optimization problem to be solved is highly non-convex due to the non-convex objectives as well as the separation of the set of feasible entry paths into several disconnected regions by the bone structures. As a consequence, many local minima may exist which are undesirable treatment options, as they are considerably inferior to the global optimum. The used adapted hyperboxing method utilizes gradient-based solver routines which only guaranteed to converge to a local minima. Hence, in order to approximate the Pareto front as good as possible, the optimization problems should be run from several promising starting points, i.e. access paths that are already close to an optimum. To determine these starting points, a projection-based brute-force optimization approach based on the method presented in Section 4.4 is used. Hence, for the starting point computation a simplified optimization problem is considered. In contrast to the overall method, a fixed target point given by the center of the target volume is assumed. Furthermore, it does not incorporate the thermal ablative effectiveness. Hence, only \( f_1 \ldots f_4 \) are taken into account. The main differences to the method presented in Section 4.4 are:

- **Objective** \( f_3 \): Contains the unsigned latitude.

- **Objective** \( f_4 \): Contains the difference of the longitude to a preferred longitude. This has been set to \( \frac{3}{4} \pi \), which corresponds to an access from 45° right ventral.

- **Scalarization**: The weighted sum method is used for scalarization.

- **Normalization**: The weighted sum method requires a transformation of the objectives to comparable ranges [132] similar to the individual utility functions discussed in Section 4.4.2. However, here, a modeling of the clinical expertise is not required, since the preference is articulated posteriori in the navigation stage. Instead, a linear normalization [132] is applied. Still, the normalization must distinguish between two cases: first, low values are in
4.5. Pareto-Navigation for Optimization of Access Paths

general better than high values \((f_1, f_3, f_4)\):

\[
f_i^\sim(x) = \frac{f_i(x) - f_i^{\min}}{f_i^{\max} - f_i^{\min}} - 1 \tag{4.17}
\]

and, second, high values are in general better than low values \((f_2)\):

\[
f_i^\sim(x) = -\frac{f_i(x) - f_i^{\min}}{f_i^{\max} - f_i^{\min}} \tag{4.18}
\]

Hence, preferred values are mapped to \(-1\) while unwanted values are mapped to 0. Due to the brute-force approach, all possible objective values for the case at hand are known. Hence, the \(f_i^{\max}\) and \(f_i^{\min}\) can be automatically derived from the objective maps.

- **Constraint:** In addition to the limitation of \(\phi\) to \([-\pi/3, \pi/3]\), the feasible set is constrained by invalid paths, i.e. paths that have a value of 0 for at least one of the normalized objectives. Based on the normalized objective maps, a respective constraint map is generated and used to mask the result of the weighted sum:

\[
c(x) = \prod_{i=1}^{4} c_i(x), \quad c_i(x) = \begin{cases} 
0 & \text{if } f_i^\sim(x) = 0 \\
1 & \text{otherwise}
\end{cases} \tag{4.19}
\]

- **Optimum extraction:** In contrast to the method proposed in Section 4.4, the minimum is extracted:

\[
f(x) := c(x) \sum_{i=1}^{4} f_i^\sim(x)w_i \rightarrow \min \tag{4.20}
\]

This modified projection-based approach is used in an iterative manner to sample the Pareto front (multiple run approach, see Section 2.5). The projection, the normalization and the constraint extraction are carried out once. However, the computation of the weighted sum of the objective maps and the consecutive extraction of the minimum are carried out for a discrete number of weighting factor sets. Each of the four weighting factors may take one of \(n\) evenly distributed values from a given range (e.g., \([0.01, 1.0]\)). For each of the possible \(n^4\) permutations, the weighted sum of the normalized objective maps is computed. Next, the constraint map is applied to the weighted sum and the pixels with the lowest resulting value are extracted. The pixel row and column correspond to latitude and longitude angles thus defining paths together with the target volume center. The paths are stored into a list. After the weighting factor set iteration, duplicates are deleted from the list because different weighting factor combinations might lead to the same minima. The overall method for seed path computation is summarized in Figure 4.15.
4.5.4 Objective Functions

So-called objective functions are used to calculate the clinical criteria for an access path during optimization. These methods are called by the optimizer using a candidate access path \( x \) defined by target point and path orientation as input. In addition to the actual objective values, the optimizer also requests the gradients of the objectives. For this purpose, a central difference operator with appropriate step sizes is used.

**Penetration depth** \((f_1)\): The penetration depth is computed using image processing. The mask of the patient is sampled along the line that corresponds to the path \( x \). The distance of the target point to the first sample that is classified as being outside the patient is returned.

**Distance to the closest risk structures** \((f_2)\): This is computed similar to \( f_1 \). The pre-computed distance map (see Section 4.5.1) is sampled along the path. The minimum of these samples is returned.

**Angulation** \((f_3\) and \(f_4)\): The angulation to the transverse plane \((f_3)\) as well as the angle in the transverse plane \((f_4)\) constitute simple arithmetic calculations. The latter is computed as the angle between the orientation projected onto the transverse plane and a preferred direction in that plane (e.g., 45° right ventral).

**Thermal ablative effectiveness** \((f_5)\): The thermal ablative effectiveness of a candidate path can be calculated utilizing various methods. Complex mathematical
4.5. Pareto-Navigation for Optimization of Access Paths

Simulations may be used to compute the heat distribution under consideration of the cooling effects of surrounding vascular structures. The resulting temperature is calculated via the electrostatic as well as Pennes’ bioheat transfer equation [117]. However, since computing those therapy outcomes is very time consuming due to the mathematical complexity, the ablation zone approximation method of Rieder et al. [12] is used instead. This method is based on two scalar fields, one representing the distance to the electrode, the other the perfusion resulting from vascular structures. The latter can be precomputed once as it is completely independent of the placement of the instrument. The published method derives visualizations from a combination of both fields. Here, this combined field, which is called ablative effectiveness field in the following, is directly used. The field does not contain real temperatures. Instead, relative values between 0 and 1 are output, where 1 corresponds to maximum achievable temperature, and, 0 to no heating relative to the normal body temperature. Hence, the mean value of the ablative effectiveness field inside the tumor mask, which is in the range \([0, 1]\), has to be maximized. Figure 4.16 shows the ablative effectiveness field for an example configuration.

**Normalization:** Before being used by the numerical optimization, the objective values are normalized by linear mapping to a range between \(-1\) and 0, with \(-1\) and 0 denoting optimal and poor fulfillment, respectively. Table 4.2 lists the ranges of the five objectives that are mapped to this output range. Values outside these ranges are clamped accordingly before the mapping. However, the ranges are chosen such that the possible objective values are covered in most cases.

![Figure 4.16](image-url)
Table 4.2: Ranges of the five objectives as used for the normalization.

<table>
<thead>
<tr>
<th>Objective</th>
<th>Poor (→ 0)</th>
<th>Optimal (→ −1)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Penetration Depth $f_1$</td>
<td>250 mm</td>
<td>40 mm</td>
</tr>
<tr>
<td>Risk Structure Distance $f_2$</td>
<td>0 mm</td>
<td>10 mm</td>
</tr>
<tr>
<td>Angulation to Plane $f_3$</td>
<td>60°</td>
<td>0°</td>
</tr>
<tr>
<td>Angulation in Plane $f_4$</td>
<td>180°</td>
<td>0°</td>
</tr>
<tr>
<td>Thermal ablative effectiveness $f_5$</td>
<td>0</td>
<td>1</td>
</tr>
</tbody>
</table>

Objectives and constraints: The result may be interpreted as an objective or constraint by the optimizer (also called soft and strict constraints respectively in the literature [38]): for an objective, all values are considered acceptable while constraints returning a value of 0 would render the candidate path invalid. Most of the described evaluation functions are interpreted as objectives. Only the distance to critical structures is used as a constraint in order to prevent penetration of bones or vessels.

4.5.5 Results
The proposed method is implemented using the rapid prototyping platform MeVisLab [164, 139]. The numerical optimization and path interpolation are integrated as external libraries (Fraunhofer ITWM, Kaiserslautern, Germany). The numerical simulation [117] as well as the approximation [12] of the ablation zone already exist as MeVisLab components. The method is integrated in the Software Assistant For Interventional Radiology (SAFIR) [200, 162] (see Figure 4.18). To prepare a patient data set for multi-criteria optimization, the software provides semi-automatic methods for segmentation of tumors [140] and vessels [12] in CT images. The patient, ribs and lungs are automatically extracted from the CT data based on thresholding.

Interactive Navigation For the interactive exploration of the solution set, an implementation of the navigation concept described by Küfer et al. [118] is integrated. The GUI contains for each objective a Pareto slider, a vertical slider with additional controls. A selector handle allows to select the aspired value for one objective within the range that is indicated by the legend. That range starts with the best possible value for an objective at the bottom, while the upper end represents the worst possible objective value existing in the set of solutions. Hence, the closer the selector is moved to the bottom of the slider, the higher the aspired quality for the objective. The impact of moving the slider of one objective by a certain amount of the available slider range varies depending on the case at hand. Also, the sensitivity of the various sliders differs since they represent different independent objectives. Modification of any selector leads to the computation of a solution that fulfills the aspired value. All other selectors are immediately adapted according to the found solution. An
4.5. Pareto-Navigation for Optimization of Access Paths

Restrictor handle
Selector handle
Lower bound

(a)

(b)

Figure 4.17: Illustration of two Pareto sliders (a) and a 2D Pareto front (b). Compare selector and restrictor positions for objective \( f_1 \) with the Pareto front and the resulting Pareto slider of \( f_2 \): the Pareto front is restricted (dotted line) which is represented by the lower bound of \( f_2 \).

The result of the interactive selection is the corresponding access path which is displayed as a virtual instrument in the context of a 2D or 3D visualization of the patient’s anatomy (see Figure 4.18). Continuous adaption of a Pareto slider typically leads to continuous movement of the instrument according to the interpolation of the Pareto efficient paths that have been determined during the optimization. However, the Pareto front has discontinuities resulting from the rib cage polytopes. Hence, manipulation of the Pareto sliders might also result in discontinuities of the applicator movement as the instrument "jumps" from one intercostal space to another in order to realize the aspired objective value.

4.5.6 Evaluation

A retrospective evaluation study is conducted with two highly experienced interventional radiologists (R1 and R2). The goal is to assess the appropriateness of the developed method for the identification of meaningful therapy strategies and to evaluate to what extent the clinicians trust the results.

Contrast-enhanced CT image data sets (venous phase, 1 to 2 mm slice thickness, 512 x 512 in-plane resolution, 204 – 761 slices) from 19 RF ablations in the liver are used to define access paths retrospectively. The segmentation, the preprocessing and the optimization itself are carried out in SAFIR for all cases prior to the actual study. The study focuses only on the exploration of the optimization results using the
Figure 4.18: Screenshot of the navigation widget inside SAFIR: The user interactively adapts the target values for the optimization objectives using the Pareto sliders. If, for example, the slider representing the distance to critical structures ("Distance", second slider) is moved down (i.e., selecting a solution with higher distance), all other sliders and the virtual instrument immediately change according to the updated access path proposal. As a consequence of the new access path, the vertical angulation ("Angulation", third slider) as well as in-plane orientation ("In-Plane", fourth slider) and thermal ablative effectiveness ("Effect", fifth slider) objectives are impaired for this specific case. As a positive side effect, the penetration depth ("Depth", first slider) is improved.

interactive navigation GUI and is performed using a study tool developed specifically for this purpose. During the study, both radiologists have to determine an access path using the interactive trade-off definition for each data set. The duration of the performed interaction is measured. The resulting paths are rated using a four-point Likert scale as either optimal, ok, unsure or bad (see Table 4.3). Both participants are able to define satisfying (choices optimal and ok) access paths in an acceptable time in 74% and 95% of the cases. As a fall back solution, the manual definition of a path is possible. In 42% and 32% of the cases, the clinicians are able to define better access paths manually. The durations of the interactive trade-off definition and the alternative manual planning are summarized in Figure 4.19 (a). The plots indicate a comparable duration for both approaches in the range of 2 to 3 min per case.
4.5. Pareto-Navigation for Optimization of Access Paths

Quality Manual Paths

<table>
<thead>
<tr>
<th>Quality</th>
<th>#Optimal</th>
<th>#Ok</th>
<th>#Unsure</th>
<th>#Bad</th>
<th>#Better</th>
<th>#Similar</th>
</tr>
</thead>
<tbody>
<tr>
<td>R1</td>
<td>10</td>
<td>4</td>
<td>1</td>
<td>4</td>
<td>8</td>
<td>8</td>
</tr>
<tr>
<td>R2</td>
<td>8</td>
<td>10</td>
<td>0</td>
<td>1</td>
<td>9</td>
<td>6</td>
</tr>
</tbody>
</table>

Table 4.3: Results of the retrospective study: For the 19 cases, the clinicians assign one of four quality categories to the paths resulting from the navigation (columns 2 – 5 list the number of cases assigned to the categories). If paths are defined manually in addition (number of cases in column 6), they have to decide if the manually defined path is worse, better or equal. Since the manually defined paths are at least equal in all cases, there is no column for worse.

Usability & Usage To assess the general usability of the proposed system, the clinicians are asked to fill out the User Experience Questionnaire [119] (see also Section 3.4.11). The results for the scales attractiveness, perspicuity, efficiency, dependability, stimulation and novelty are 1.583, 1.5, 1.375, 1.0, 2.0, and 1.5 respectively. Although this test is aimed at a higher number of participants, these results indicate a good usability already.

In addition, questions regarding specific aspects of the system have to be answered. Both participants judge the distance to critical structures and the thermal ablative effectiveness as the most important objective, followed by the vertical angulation and the penetration depth. This is also reflected in the order and frequency of usage of the associated Pareto sliders. In-plane orientation is mainly modified to exclude dorsal approaches. Both participants appraise the multi-objective approach as reasonable for RFA treatment planning and think, that it may help to define therapy alternatives that might not be found with manual planning, or might at least facilitate faster therapy definition. They judge the benefit to be higher for less experienced clinicians.

Performance The performance of the method is measured during the retrospective study as well (Core i7 3.33 Ghz, NVIDIA GeForce GTX 295). While the duration of preprocessing and optimization is in the range of minutes, the navigation using the proposed GUI can be carried out interactively without any noticeable delay. The runtime of all preprocessing and optimization steps is summarized in Figure 4.19 (b). The median duration of the Euclidean distance transform, the starting point computation, the generation of the polytopes, and the calculation of the vascular perfusion field for the approximation of the thermal ablative effectiveness computation together is 76 s, while the median duration of the Pareto front approximation including evaluation function computations is 811 s. Hence, the Pareto front approximation is roughly as 10 times as expensive as all other steps combined. Strong outliers
Figure 4.19: Plot (a) summarizes the durations (vertical axis, seconds) of access path determination using the interactive trade-off definition approach (left) and the alternative manual planning (right) for both clinicians (R1 and R2). The durations (vertical axis, given in seconds) needed for starting point computation (P1), generation of the polytopes (P2), calculation of vascular perfusion for the approximation of the thermal ablative effectiveness (P3), the Euclidean distance transform (P4), and the Pareto front approximation process (Solver) are given in plot (b).

can be observed in the duration of the polytopes calculation process as well as the Pareto front approximation run-time. An analysis of the measured data reveals that the solver performance strongly depends on the number of polytopes. Exemplary, an outlier case has is re-calculated with a reduced number of polytopes. For this case, the original number of polytope points of the rib cage is 18000, whereas the average number of all cases is 7600. The significantly higher number is a result of an inaccurate rib cage mask which also includes the kidneys. After manual adjustment of the mask, the number of polytope points decreased to 5056 resulting in a reduction of the Pareto front approximation run time from 5798 s to 247 s.
4.5.7 Discussion

To the best of the author’s knowledge, the proposed method is the first multi-criteria optimization approach for RFA that utilizes a realistic representation of the thermal necrosis including cooling effects while keeping computation times in a practical range (∼15 minutes). The approximation has been shown to be comparable with a numerical simulation [12]. Hence, its utilization during optimization represents an improvement over the commonly used ellipsoids.

Another important aspect of the presented work is the utilization of Pareto optimality. This has been used for optimization of RFA planning before by Seitel et al. [172]. Major differences to their work is the incorporation of the target volume coverage and the result exploration: The presented method favors interactive objective-driven trade-off definition using the Pareto sliders, whereas the approach by Seitel et al. allows for manual selection from a list of Pareto-efficient paths. While the clinicians adjudged the interactive trade-off definition presented in this work as suited, they proposed to extend the system such that the Pareto sliders would be adapted if an access path is modified manually.

For exploration of all interesting treatment options, a global approach is necessary. To this end, a starting point generation is proposed which approximates optimal solutions for all possible weight distributions. The results depend on the anatomical situation and might vary heavily between cases. For example, multiple starting points may be in one intercostal space, while in other intercostal spaces, no starting points may be found, simply because there are no possible paths, independent on the chosen weight factor combination. Hence, starting points are placed in a way to cover interesting regions. Together with differentiable objective and constraints functions, this enables the utilization of a local gradient descent strategy as a second step.

The preliminary study presented in this work is a first step to evaluate the proposed method. Planning using the approach takes around 3 min per case which is comparable to the manual planning. Furthermore, this is in the same range reported by Seitel et al. Using the presented approach, satisfying treatment plans could be generated in more than 75% of the cases. Still, in almost 50% the clinicians preferred to define access paths manually. However, it is possible, that the clinicians stopped the Pareto navigation while having selected an unsatisfying path although a better one had been reached before during navigation. An intuitive way to recreate previous results is however missing, as moving one Pareto slider back to a previous position does not necessarily recreate the previous situation. The reason for that is that multiple solutions might have a similar value for that objective. Hence, there is still potential for improvement of the navigation process. Nonetheless, these first results are promising, especially given that the involved clinicians are highly experienced in planning of such procedures. Hence, it can be assumed that planning carried out by less experienced radiologists could benefit even more from the proposed workflow. This is also supported by statements of the clinicians.

A strong dependency of the quality of planning results to the segmentation quality
was observed during the study. Many of the problematic cases resulted at least partially from poor segmentation results. While smaller errors might not have measurable effects, missing representation of structures may lead to unfeasible results. Hence, a more robust and if possible largely automatic segmentation pipeline would be beneficial. Objectives that would require a liver mask \(f_6, f_7\) are not considered by the implementation of the method as presented here. The inclusion of such criteria is, however, straight forward, if such a segmentation is available.

Blood vessels are considered during seed path generation and criteria evaluation. However, during the Pareto front approximation, they are not considered in the form of polytopes since this would raise the numerical complexity immensely. Although penetration of vessels during interaction has been reported only once during the study, a good trade-off might be to represent large vascular structures also as polytopes. Except the utilization of the rib cage polytopes for the restriction of the feasible set, the proposed method mainly operates directly on the input segmentation masks. Hence, computation time, errors and parameters of according preprocessing steps are reduced. In contrast, most publications from the field completely rely on mesh representations. However, a more robust and adaptive generation of the polytopes is desirable, as the number of polytopes influences the performance strongly. Depending on the clinical workflow, therapy planning might be performed shortly before the actual intervention. In that case, the average computation duration of the method (~15 minutes) might still be too long. In addition to the optimization of the polytope generation, the performance of the solver could be improved by parallelization. Especially a patch-wise subdivision of the problem seems promising.

4.6 Discussion and Conclusion

In this chapter, two methods for the multi-objective optimization (MOO) of access paths in the context of tissue ablation planning are presented. Both methods aim at the incorporation of multiple clinically motivated criteria as described in Section 2.2.2. Although both methods have been primarily developed for the support of the planning process of RFA in the liver, the adaption to other modalities and target organs is possible and mainly involves the adaption of the used objectives. As shown in Section 4.4.5, the first method has already been evaluated for the planning of liver biopsy.

Both developed methods use a similar set of objectives. However, not the same anatomical structures have been taken into account. The first method does not use masks of hepatic vascular structures because a simple-to-use method was not available at the time of development. The second method does not use a liver mask because the segmentation algorithm used for the first method was developed by a project partner and could only be used in that single project.

The first method (see Section 4.4) computes a list of path proposals for a given target point inside the liver based on a weighted combination of cylindrical projections. Each projection represents one objective for all possible paths and is either generated...
using the GPU or based on simple arithmetic calculations. Given that the target point independent preprocessing step has been carried out (which takes around 5 s), this brute force approach allows to generate the list of path proposals in less than one second and therefore enables the iterative manual refinement of the target point. In contrast to most of the works from this field, the proposed approach directly operates on segmentation masks. The conversion of anatomical structures to other representations is not necessary. One current drawback of the projection-based approach is that the target volume coverage cannot be treated in a realistic manner. Furthermore, the result exploration aims at a minimalistic interaction. The iteration over three proposals allows for operator-independent and reproducible results. In that way, the method could be integrated into a planning software in a very decent way. The result could be simply used as a default placement when the user wants to add a new instrument since it is so fast and requires no interaction. However, some users may prefer to have more influence on the result selection. To overcome these two major points of criticism, a second method has been developed. It uses the previous projection based approach to compute starting points for a numerical optimization. Since this optimization evaluates single paths individually, it allows to incorporate a realistic representation of the probable ablation zone. To this end, a fast approximation using the GPU is used which allows to keep computation times in a range that is still practical. Furthermore, the utilized optimization approach, which is based on an approximation of the Pareto front, allows to interactively explore access path alternatives based on the definition of aspired values for the objectives using a slider based GUI. Both methods have been evaluated in clinical studies. Overall, the quality of the achieved access paths has been better for the first method. Major reasons for unsatisfying results of the second method are defects in the used segmentation results and the unreproducible nature of the navigation paradigm. Method one is more robust with regard to the quality and completeness of the segmentation results. Furthermore, it requires far less computation time (6 s vs. 15 min). Nonetheless, the incorporation of the realistic ablation zone approximation as well as objective-driven interactive exploration of the Pareto front are desirable characteristics. Hence, they should be either translated to the first method, or the robustness of the second method should be improved. Further work will concentrate on the planning of tissue ablation for larger tumors, which requires the optimal placement of multiple instruments. Existing methods in this field use strong simplifications such as ellipsoidal ablation zones and predefined skin entry points [187, 156, 155].
Chapter 4. Multi-Objective Optimization of Access Paths for Percutaneous Tissue Ablation

4.7 Outlook

First steps have been done to advance the method presented in Section 4.4 according to the discussion above. The following aspects have to be addressed to adapt the positive characteristics of the second method (Section 4.5):

- Consider the whole target volume instead of a fixed target point.
- Allow for interactive exploration of access paths.
- Create an objective map describing the ablative effectiveness in a realistic manner.

The target volume can be incorporated by applying the method to multiple points inside the volume. To this end, a regular grid is used. However, other sampling patterns are also possible. The proposed method for optima extraction can simply be applied to each sample point and the lists of optima joined and sorted afterwards. To advance towards an interactive exploration of the data, it can be interpreted as a multivariate dataset, if the objectives are considered to be separate dimensions. Hence, established methods for the analysis of multivariate datasets can be used. The concept of parallel coordinates [94] in particular bears many similarities to the GUI used for the exploration of the Pareto front. Hence, each of the defined objectives may define one vertical axis in the parallel coordinates plot. The ranges of the axes are scaled such that all axes cover the same vertical space (see Figure 4.20). The axes are inverted for objectives which favor higher values over low values. Among the described objectives this only applies to the distance to risk structures.

Each path is visualized as a line sequence intersecting each axis at the point that corresponds to the path’s value for the objective that is represented by the axis. To enable visual distinction of paths even for large numbers of paths, the brightness and opacity of the lines is selected depending on the objective values, i.e. lines with preferred objective values are displayed brighter than lines with unwanted values.

The initial display of the parallel coordinates plot contains all paths. To reduce the

Figure 4.20: Illustration of the representation of access paths in the parallel coordinates plot: Path 1 represents an ideal path with good values for all objectives. Path 2 exhibits good and bad properties while path 3 features unwanted values for all objectives.
Figure 4.21: The initial parallel coordinates plot (a) is filtered using restrictor sliders (b): The insertion depth is restricted to a maximum of 150 mm (leftmost axis), the distance to risk structures to a minimum of 4 mm (second axis from the left) and the vertical angulation to a maximum of $10^\circ$ (second axis from the right). The selected path (red line) is visualized in a 3D rendering together with the anatomy of the patient (c).

number of displayed paths, restrictor sliders on the axes are used to select ranges of the objectives (see Figure 4.21 (a, b)). Lines outside of the defined ranges disappear. Thus, the set of displayed paths can be quickly reduced to the interesting candidates. The remaining paths can be explored by simply moving the mouse over the plot: the line below the mouse cursor is automatically highlighted. The respective path is visualized in real-time in a 3D view together with a rendering of the anatomy. A click with the left mouse button on the plot either locks or unlocks the selection. To enable the selection of Pareto optimal paths only, this approach should be combined with the weighting factor set iteration as described in Section 4.5.3. The result would represent a compromise between the method proposed by Seitel et al. [172] and the method proposed in Section 4.5.

The incorporation of a realistic estimation of the ablation zone represents the most challenging aspect as this cannot be done during the projection. A possible solution is the computation of the corresponding objective value for a small subset of trajectories in the equirectangular coordinate system and to interpolate the remaining values. The approximation of the simulation proposed by Rieder et al. [12] allows to compute a single ablation zone in a duration far below 1 s. Hence, a reasonable sampling of the feasible set in a practicable computation time seems realistic.
Chapter 4. Multi-Objective Optimization of Access Paths for Percutaneous Tissue Ablation

Contributions and Publications  The first method described in this chapter (Section 4.4) was presented at the SPIE Medical Imaging conference 2010 in San Diego, USA [17]:


Study 1 was presented at the 91. Deutscher Röntgenkongress (DRK) 2010 in Berlin, Germany [14] and at the European Congress of Radiology (ECR) 2012 in Vienna, Austria [15]:


Study 2 was published in 2016 in the International Journal of Computer Assisted Radiology and Surgery (IJCARS) [8]:


The proposed combination of this method with an interactive exploration utilizing parallel coordinates has been presented at the 12th Annual Meeting of the German Society of Computer and Robot-assisted Surgery (CURAC), Innsbruck, Austria [19]:

The second method (Section 4.5) including the accompanying evaluation study was presented at the 6th International Conference on Information Processing in Computer-Assisted Interventions (IPCAI) 2015 in Barcelona, Spain and published in the International Journal of Computer Assisted Radiology and Surgery (IJCARS) [21]:


The development of this method was a team effort with fundamental contributions from colleagues at Fraunhofer MEVIS in Bremen and Fraunhofer ITWM in Kaiserслаutern. The overall framework that integrated the image processing based methods together with the numerical approach was developed by the whole team. The numerical optimization and the interpolation of the Pareto front were developed completely at Fraunhofer ITWM by Philipp Süss and Katrin Teichert and are partially based on the PhD Thesis of Katrin Teichert [184]. Sabrina Haase (Fraunhofer MEVIS) integrated numerical simulation models for the prediction of the ablation zone. Christian Rieder (Fraunhofer MEVIS) integrated the according approximation and was also responsible for the implementation of the graphical user interface for the interactive exploration.
5 Interactive Visualization of High-Intensity Focused Ultrasound Sonications

My God man, drilling holes in his head is not the answer!

(Dr. McCoy, Star Trek IV - The Voyage Home)

Planning of high-intensity focused ultrasound (HIFU) involves the definition of multiple sonications in order to cover the target volume, where each sonication leads to the thermal ablation of a more or less elliptic volume. However, the exact position, size and shape of the ablation zones is not known in general. These properties are influenced by characteristics of the transducer and the tissues encountered along the beam path. Numerical simulation is an established tool to predict the heating and the induced damage resulting from HIFU. However, the required computations are quite complex and time consuming and do not allow for immediate feedback during interactive sonication planning. In this chapter, a method for interactive approximation of the temperature and thermal dose of HIFU sonications based on interpolation of profiles derived from numerical simulation is presented. Multiple sonications can be displayed in real-time allowing for the interactive determination of a therapy that covers the whole target volume.
Chapter 5. Interactive Visualization of High-Intensity Focused Ultrasound Sonications

5.1 Introduction

High-intensity focused ultrasound (HIFU) offers a promising tool for non-invasive thermal treatment of benign and malignant changes of soft tissue in the human body. During the planning of such interventions, a strategy has to be defined which allows for the coverage of a target volume by many small individual ablation zones. This is known as the point-by-point method. During planning, an estimation of the damage resulting from the individual sonications is necessary to assess the quality of the plan. However, the exact position, size, orientation, and shape of the ablation zones depends on many factors (see Section 2.3.2). Typically, numerical simulation is used to predict the resulting heating, and, if needed, the thermal damage. However, the computational complexity is very high since expensive computations on highly resolved meshes are required. Hence, the result of each plan modification can only be reviewed after a duration of several minutes or even hours. Recent developments improve the execution performance greatly by utilizing the computational power of the GPU [4]. Depending on the number of sonications of a treatment plan, the computation duration is now in a range of a few seconds. Although this allows for fast prediction of the therapeutic effect resulting from a therapy plan, it does not facilitate interactive planning. For that, interactive approximations are required that predict the outcome of a sonication simulation reasonably well. For the planning of needle-based thermal ablation procedures, various according methods have been proposed, which allow for interactive ablation strategy determination based on a realistic prediction of the ablation outcome including the heat sink effect (see Section 4.2). However, no comparable approximation approach has been devised for the planning of HIFU yet.

In this chapter, a novel method for interactive approximation of the sonication resulting from HIFU sonications in the liver is presented. It is based on the patient independent sampling of actual numerical simulation results for various configurations of the HIFU setting. Heterogeneous tissue, except high attenuation structures such as ribs, is considered. This patient independent step has to be executed only once for a given transducer. The result are lookup tables containing parameters and profiles describing the temperature and thermal dose distributions, which can be reused during interactive sonication planning. Multivariate interpolation is used to reconstruct corresponding results for individual sonications in real time. Heat sink effects are estimated by means of a thermal equilibrium representation which is computed based on the patient specific vasculature in the region to be treated.

This chapter is structured as follows: Section 5.2 reviews related work from the field of simulation and approximation of sonication formation in thermal ablation procedures. In Section 5.3, the method will be presented in detail. After results are presented in Section 5.4, the chapter closes with a discussion of the presented work (Section 5.5).
5.2 Related Work

Various treatment planning systems have been proposed by several groups to support the process of sonication planning [35, 128, 3]. They follow a common outline quite similar to planning systems for needle-based interventions: patient images are imported and important structures are segmented. The segmented structures form the input for numerical simulation algorithms that predict the heating resulting from the user defined therapy.

The simulation of transcutaneous HIFU incorporates the modeling of the propagation of the ultrasound wave and the heat diffusion in the tissue. Furthermore, the damage can be computed based on the temperature and the treatment time using a thermal dose model (see Section 2.1.3). Typically, the propagation of the ultrasound wave resulting from phased array transducers is modeled using the time-harmonic Helmholtz equation [97, 4, 99]. Respective numerical methods are however computationally expensive due to the high resolution meshes involved [4]. If only homogeneous material is considered, several simplifications are possible. Using the Rayleigh-Sommerfeld integral, the pressure fields of single transducer elements can be described [4]. The angular spectrum method can be used to efficiently propagate the acoustic wave in a slice-wise manner through the computational domain [99]. A combination of this approach with an additional phase correction yields the hybrid angular spectrum method which also accounts for material inhomogeneities [198].

Georgii et al. [4] propose a HIFU-simulation method which runs on the GPU. Their overall approach is based on the separation into a homogeneous and a heterogeneous domain outside and inside the body, respectively. In the homogeneous domain, the pressure field is composed of the Rayleigh-Sommerfeld integrals of the individual elements. Inside the body, the hybrid angular spectrum method is used. This sound wave propagation model is very efficient and needs to be computed only once, independent of the sonication duration. It is combined with a subsequent computation of the bioheat transfer equation. The authors present implementations of the method on the CPU and GPU and report for the simulation of a single sonication on a $256^3$ grid computation times of approximately 400 s (Intel® Core™i7, 3.0 GHz) and 3 s (NVIDIA GeForce® GTX 480), respectively.

Schwenke et al. [171] present an atlas-based approximation of the pressure field for the simulation of HIFU treatments in the liver during respiratory motion. In this setting, the ultrasound pressure field needs to be updated in each time step due to the moving liver, which increases the computational efforts considerably compared to the static case. They propose the utilization of an atlas which stores precomputed pressure fields for potential target points on a grid with a spacing of 5 mm. The simulations are carried out for a power of 1 W and in a homogeneous medium. For the approximation of the pressure field for a selected sonication, the pressure field for the closest grid point is used and the amplitude scaled according to the used power. The field is translated in order to compensate the distance to the grid point and transferred to the GPU, where it is further adapted to account for heterogeneous
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attenuation. This is achieved by splitting the overall attenuation into a homogeneous component and a local deviation from the homogeneous absorption. Similar to the hybrid angular spectrum method, a sweeping algorithm is used to accumulate and propagate the deviations from slice to slice. Shadowing resulting from the ribs is also computed in a slice-by-slice fashion. The acoustic pressure in shadowed regions is reduced to 10% of the original value. By distributing the remaining power to unshadowed regions, the overall acoustic power is conserved. As a result, the method models the higher energy deposition of the smaller beam through the intercostal space. The pressure field approximation is carried out in less than 50 ms. Its results are used as the input for the GPU-based computation of the heat, which requires ~20 ms per time step. Hence, depending on the size of the time steps, the simulation duration can be shorter than the duration of the simulated treatment. Neufeld et al. [145] report a general framework for the simulation of thermal ablation procedures including RFA, MWA, and HIFU. Their implementation also uses the GPU to accelerate computations. However, only few details are given in the publication. The computational duration is not stated at all.

5.3 Approximation and Visualization of Sonications for Planning of High-Intensity Focused Ultrasound

The proposed method does not aim at the advancement of the numerical simulation methodology. Instead, it treats the numerical simulation as a black box and stores its result in a manner that allows to reuse and combine them during interactive planning. It follows ideas that are similar to those presented by Kröger et al. [116], Rieder et al. [12] (see Section 4.2) and Schwenke et al. [171].

In the following, aspects that influence the ablation formation during HIFU are described and the proposed method is outlined. In the subsequent sections, the various stages of the algorithm are described in detail.

5.3.1 Method overview

The basic idea of the method is the separation of influences on the sonication formation into various classes:

- **Basic sonication formation**: the formation of the sonication depends on the relative position of the focal spot in relation to the transducer as well as transducer settings such as sonication power and duration.

- **Material layer dependencies**: the sonication formation is further influenced by tissue layers which are typically encountered in the course of the beam path.
5.3. Approximation and Visualization of Sonications for Planning of High-Intensity Focused Ultrasound

• **Heat sink effect**: cooling of nearby vessels leads to modification of the sonication.

• **High attenuation media**: bones and other structures with a strong acoustic absorption may influence the sonication formation further. Additionally, unwanted hot spots might form at the interfaces of such structures.

The proposed method considers the first three phenomena. It aims at a systematic analysis of these aspects for a number of representative example configurations and uses the results for a prediction of the temperature or thermal dose distribution in a patient individual case. The algorithm consists of the following three major stages:

• **Preprocessing (transducer model specific)**: The first step is a time consuming preprocessing which has to be executed once for every transducer model. It constructs two lookup tables which describe ablation zones with regard to temperature and thermal dose. Therefore, for a limited number of combinations of parameters that influence the shape of a sonication, such as position in space, transducer settings as well as the layers of tissue in the beam path, compact descriptions of the sonication are computed based on numerical simulation.

• **Initialization (transducer placement specific)**: The placement of the transducer determines the thickness of tissue layers encountered by the beam path. Hence, reduced lookup tables are constructed for the tissue layer configuration at hand in order to reduce the number of dimensions that need to be considered during sonication definition. Furthermore, the placement restricts the possible domain for the planning of sonications. Inside this domain, a field describing the perfusion of the local vasculature is computed.

• **Visualization (sonication specific)**: The compact sonication description is interpolated from the lookup tables for the sonication at hand. The result is combined with the precomputed perfusion field in order to approximate the heat sink effect.

The algorithm treats temperature and thermal dose in exactly the same way. All computations are carried out in parallel for both measures. In the following, the term *energy* is used to refer to both measures.

### 5.3.2 Transducer-Specific Preprocessing

For a given transducer, the energy distributions for a representative subset of all possible configurations have to be sampled. This can be compared with a large number of phantom experiments: to determine the effect of the depth of the focal spot inside the tissue, its position is changed multiple times, and the results of according HIFU experiments are measured. The same can be done for the duration
and power of the sonication. And, in order to evaluate the effect of various tissues, the composition of the phantom itself may be changed. Here, these experiments are carried out using a virtual phantom and numerical simulation. For that, iterations over multiple dimensions representing the parameters of an experiment configuration have to be carried out. Before this iterative process is described, the utilized media layer model and the necessary parameter space dimensions are introduced.

**A simplified media layer model:** A representation of the virtual phantom is needed which is represented by as few parameters as possible. The targeted treatment scenario is tumor ablation in the liver. Hence, the typical media encountered by the HIFU beam in a hepatic treatment is modeled as a fixed sequence of interfaces: first, outside the body, the beam travels through degassed water, then through adipose and finally through liver tissue. Furthermore, a simplified geometry is assumed. It consists of parallel, planar media interfaces which are placed perpendicular to the main axis of the transducer (see Figure 5.1 (a)). Configurations of the resulting media layer model can be described using two scalar parameters: the depths $d_w$ and $d_a$ of the water and adipose layers, respectively.

**Description of sonication setting:** Considering the media layer model and the sonication related parameters described before, the properties of one experiment configuration are as follows:

- Focal spot $p = (p_x, p_y, p_z)^T \in \mathbb{R}^3$: the relative position of the focal spot in relation to the transducer coordinate system.

![Figure 5.1](image.png)

Figure 5.1: Image (a) shows the transducer coordinate system including the simplified media layer model and an exemplary sonication reference frame. Image (b) illustrates an sonication description which consists of the reference frame (top) and an energy profile (lower).
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- Sonication power $P \in \mathbb{R}$
- Sonication duration $t \in \mathbb{R}$
- Water layer depth $d_w \in \mathbb{R}$
- Adipose layer depth $d_a \in \mathbb{R}$

Hence, this simplified experiment setup can be described by seven scalar parameters and the resulting lookup table parameter space consists of seven dimensions.

**Seven-dimensional sampling:** Each of the resulting seven dimensions is sampled regularly. The ranges for most of the seven dimensions depend on the modeled transducer: the range of water layer depth values $d_w$ has to match the distances between transducer and body and therefore depends on the way the transducer is integrated into the patient table. The spatial parameters $p_x$, $p_y$ and $p_z$ are related to the transducer coordinate system and have to cover the volume that is typically of interest for a certain model. Likewise, power and duration ranges are transducer specific. The range of $d_a$ on the other hand should be chosen such that typical distances between skin and liver are covered.

**A compact and interpolatable HIFU sonication description:** It is assumed, that, in the absence of any heat sink effects, the energy distribution of a sonication is more or less rotationally symmetric with respect to a certain reference frame (depicted as enclosing box in Figure 5.1 (b)). Hence, the energy distribution of a sonication for the given focal spot $p \in \mathbb{R}^3$ in the coordinate system of the transducer is represented by:

- **Offset** $v_o \in \mathbb{R}^3$: the center of the sonication $p_c$ is not necessarily identical with the planned focal spot due to phase aberrations caused by tissue inhomogeneities [93]. Hence, $v_o$ represents the offset between $p$ and $p_c$.

- **Orientation** $o \in \mathbb{H}$: The orientation of the sonication reference frame with regard to the main axis of the transducer. To allow for interpolation, a quaternion\(^1\) representation is used [177].

- **Size** $s = (s_x, s_y, s_z)^T \in \mathbb{R}^3$: The extent of the reference frame.

- **Energy profile** $e \in \mathbb{R}^{u \times v}$: A two-dimensional profile containing energy values for a discrete number of two-dimensional coordinates.

\(^1\)\(\mathbb{H}\) denotes the algebra of quaternions. It is named after William Rowan Hamilton, who first described quaternions.
**Chapter 5. Interactive Visualization of High-Intensity Focused Ultrasound Sonications**

**Sonication description determination:** For each grid point in the lookup parameter space, a numerical simulation computes temperature, and thermal dose fields. For each of the two energy representations, the following algorithm is used to determine the corresponding sonication description:

1. **Restriction to initial volume of interest (VOI):** the energy volume is restricted to a cuboid region of a reasonable initial size (e.g. $40 \times 40 \times 60 \text{ mm}^3$) in order to reduce the computational complexity. The center of this VOI is the focal spot $p$.

2. **Detection of sonication in VOI:** a sonication is represented as a bright blob inside the VOI. Typically, its center is not exactly at the focal spot $p$. To detect and isolate the sonication, the local maximum closest to $p$ is searched first. This point is used as the seed for an iterative region growing approach [40] which allows to remove other blobs inside the VOI (Figure 5.2). This is especially necessary for sonications that are offset from the main transducer axis.

3. **Determination of sonication iso contours:** the fuzzy c-means algorithm [68] is applied to identify clusters of similar energy levels (see Figure 5.3 (b)). Too many levels would increase the computation time while too few levels would not allow for the following analysis steps. A number of six cluster classes represents a good compromise and is used consequently. Classes that intersect the border of the VOI are excluded: since they would be large enough to extend outside the VOI, it can be assumed that they represent very low energy levels close to the minimum value. The remaining clusters are compact enough to fit into the VOI and can be seen as iso contours of the ablation zone.

![Figure 5.2](image)

Figure 5.2: The initial VOI might contain blobs that do not belong to the sonication (a). Iterative region growing started from the local maximum *(red dot)* closest to the focal spot $p$ is used to remove those blobs (b).
5.3. Approximation and Visualization of Sonications for Planning of High-Intensity Focused Ultrasound

Figure 5.3: Determination of the sonication reference frame based on the simulation data (a): six energy levels are segmented (b). The three levels with the lowest energy extent outwards of the VOI and are therefore ignored. Ellipsoids are fitted to the remaining three levels (c-e). Only the fitting quality of the highest level is above 0.9. The orientation of the corresponding ellipsoid is used to determine the frame extent based on the lowest acceptable energy level (f).

(4) **Determination of sonication reference frame:** To each of the remaining iso contours, an ellipsoid is fitted using principal component analysis (PCA) (Figure 5.3 (c-e)). One of the ellipsoids is used to determine the sonication reference frame. The selection of the ellipsoid is based on the quality of the fittings, which is computed for each ellipsoid using the *dice coefficient*. However, there may be several fittings of comparable quality. Furthermore, the ellipsoids may have varying orientations. Hence, the selection of the ellipsoid strongly influences the orientation of the reference frame. The selection is carried out as follows: if one or more of the fittings have a dice coefficient of at least 0.9, the cluster indices of the corresponding ellipsoids are interpreted as a sorted set and the ellipsoid, which corresponds to the median index, is selected. If none of the fittings has a dice coefficient of at least 0.9, the ellipsoid with the highest dice coefficient is used. The orientation of the selected ellipsoid defines the orientation $o$ of the sonication reference frame. The center $p_c$ (and therefore the offset $v_o$) of the frame and its extent in $z$-direction is determined by intersecting the local $z$-axis of the ellipsoid with the largest remaining iso contour. Averaging the resulting two intersection points yields $p_c$. Likewise, the extent in $x$- and $y$-direction is computed by intersecting the largest remaining iso contour with the $x$- and $y$-axes of the coordinate system defined by $p_c$ and $o$ (see Figure 5.3 (f)). This approach allows for a compact reference frame, and therefore a good coverage of the relevant energy levels by the low resolution energy profile.

(5) **Energy profile computation:** The original energy field is sampled in the central $xz$- and $yz$-planes of the sonication reference frame. Both planes intersect in the local $z$-axis of the sonication. The samples in the resulting four rectangles are averaged to yield the energy profile $e$ (see Figure 5.4).
Chapter 5. Interactive Visualization of High-Intensity Focused Ultrasound Sonications

Figure 5.4: Energy profile creation: the energy values in the sonication reference frame (a) are sampled in the xz- and yz-planes (b). The values in the resulting four rectangles (c) are averaged after according mirroring (d).

The result of the sampling process is a function $f : R^7 \rightarrow R^3 \times H \times R^3 \times R^{u \times v}$:

$$f(p_x, p_y, p_z, P, t, d_w, d_a) = (v_o, o, s, e)$$

(5.1)

represented as a lookup table. This algorithm is carried out for the temperature and the thermal dose. The resulting lookup tables are $f_t$ and $f_d$, respectively.

5.3.3 Transducer Placement Specific Initialization

After the user placed the transducer with respect to the patient, several processing steps are carried out which are independent of the placement of individual sonications.

Lookup table reduction: The lookup tables depend on the two parameters $d_w$ and $d_a$. Hence, the sonication descriptions stored in the lookup table have been determined for certain depths of the water and adipose layers, respectively. These two parameters do not change during sonication placement, as long as the transducer is not repositioned. Therefore, the complexity of interpolation during sonication placement can be lowered by reducing the lookup tables based on the media layer depths resulting from the current transducer placement. These depths are first derived based on the transducer transformation matrix and a label image containing labels for water, adipose and liver. Using image processing, the distances between the transducer plane and the interface between water and adipose and the interface between adipose and liver are computed along rays parallel to the transducer axis. The water and adipose penetration depths $d_w$ and $d_a$ are computed as mean values of these distances inside a cylinder aligned along the transducer main axis. The radius of the cylinder taken into account is 50% of the transducer radius.

Using the two transducer placement specific parameters $d_w$ and $d_a$, the seven dimensional $f(p_x, p_y, p_z, P, t, d_w, d_a)$ is reduced to $f^\sim(p_x, p_y, p_z, P, t)$ by means of bilinear interpolation. For that, the components of sonication descriptions need to be interpolated individually. While the offset $v_o$, the size $s$ and the entries of the energy profile $e$ are linearly interpolated, the orientation $o$, which is given as a quaternion, is interpolated using spherical linear interpolation (slerp) [177]. The result of the reduction are the two five-dimensional lookup tables $f_t^\sim$ and $f_d^\sim$ for temperature and thermal dose, respectively.
Vessel preprocessing: The placement of the transducer defines the region of interest for sonication placement. In the clinical setting, focusing outside a certain region around the transducer main axis is not possible. Likewise, the region in which interpolation of sonication descriptions from the lookup tables is possible, is restricted. Hence, only vascular structures in this region need to be considered. The relevant volume of interest (VOI) is derived from the boundary of the space that is covered by the lookup table. However, it is aligned with the original mask image coordinate system in order to prevent the need for reformation, which might introduce errors (Figure 5.5 (a)).

The thermal equilibrium as described by Rieder et al. [12] is used to represent the cooling effects. This thermal field describes the heat transport of the vasculature in order to estimate the heat sink effects of nearby blood vessels. It is derived from a steady-state simulation of the heat transport which is computed for a given vessel mask using one very large time step (1000 s). This method is used to precompute a perfusion field \( g(x) \in \mathbb{R} \) inside the previously determined VOI for later use during the interaction sonication visualization. The values of \( g(x) \) are in the range \([0, 1]\), where 0 corresponds to no cooling. The higher \( g(x) \), the more heat is removed. If \( x \) is inside a vessel, then \( g(x) = 1 \) (Figure 5.5 (b, c)).

5.3.4 Interactive Sonication Visualization

During interactive planning, multiple sonications are placed. In the following, the process of visualizing a single sonication during the interactive manipulation of its target point \( p_t \) is described. Furthermore, the power \( P \) and the duration \( t \) can be changed at any time and therefore influence the visualization. The visualization is

![Figure 5.5](image)

Figure 5.5: After transducer placement, a VOI is extracted from the vessel mask (a). Based on this, the perfusion field is precomputed for the visualization (b and c).
implemented using a fragment shader which is applied to a rectangle representing the slice currently displayed in a 2D view.

**Sonication approximation:** The approximation of the sonication for the given parameters $p_t$, $P$, and $t$ is based on the interpolation of a sonication description for these parameters. To be able to interpolate values from $f_i\tilde{}$ or $f_d\tilde{}$, the target point $p_t$ is first transformed into the transducer domain which yields $p_t\tilde{}$. A tuple $(v_o, o, s, e)$ is interpolated from $f_i\tilde{}$ or $f_d\tilde{}$ using the parameters $p_t\tilde{}$, $P$ and $t$ using again linear and slerp interpolation. The interpolated tuple represents the sonication reference frame and the corresponding energy profile $e$. Theoretically, the sonication can be reconstructed by creating a rotationally symmetric field from $e$ and by transforming it into the reference frame. Practically, this is solved in an inverse manner inside the fragment shader for each fragment $x$:

- The position of $x$ is transformed into the reference frame coordinate system.
- It is checked if the transformed position $x\sim$ is inside the frame.
- If $x\sim$ is inside the frame, its distance $r$ to the local z-axis is computed. The energy profile $e$ is sampled using $r$ and the z-coordinate of $p\sim$ as lookup coordinates. The result is the relative energy increase $I(x)$.
- If $p\sim$ is outside the frame, the relative energy increase $I(x)$ is zero.

**Heat sink effect approximation:** The relative energy increase $I(x)$ has to be combined with the perfusion field $g(x)$. Hence, a transition function $h(g(x))$ is needed in order to yield the relative energy increase $I\sim(x)$ under perfusion:

$$I\sim(x) = I(x)h(g(x))$$  \hspace{1cm} (5.2)

An according function is determined based on simulation experiments. Consider the following three simulations:

- $s(x, P, t, p_t)$: simulation of a sonication with the power $P$ and the duration $t$ at a fixed, central position $p_t$ in the transducer domain without any cooling.
- $s_c(x, P, t, p_t)$: simulation of the same sonication, however this time under consideration of the perfusion of a synthetic vessel $v$ close to $p_t$.
- $g(x)$: perfusion computation according to Rieder et al. [12] for the vessel $v$.

$s$, $s_c$ and $g$ are clipped to a VOI encompassing the sonication. The reference frame determination is used for this. For each voxel in the VOI, the values of $s$, $g$ and $s_c$ are used as $x$, $y$, and $z$ coordinate for a point in a scatter plot. Visual inspection
5.3. Approximation and Visualization of Sonications for Planning of High-Intensity Focused Ultrasound

yields the conclusion, that the transition function can be well approximated by a Gaussian function of height 1 centered at $g = 0$:

$$I^\sim(x) = I(x) e^{-\frac{g(x)^2}{2\sigma^2}}$$

(5.3)

This is confirmed by least squares fitting of the surface given by

$$z = x e^{\frac{2}{2\sigma^2}}$$

(5.4)
to the scatter plot. This experiment is repeated for extreme values of $P$, $t$, $d_w$, $d_a$. Furthermore, $g(x)$ is shifted during these experiments to three different positions such, that the value of $g$ at the sonication center is 0.0, 0.5 and 1.0. Fitting the surface described in Equation 5.4 to the scatter plots resulting from the 48 experiments results in almost identical values of $\sigma$ for the temperature in all experiments. In the case of the thermal dose, there are slight variations. Still, in both cases a surface can be found that fits the combined scatter plots of the resulting 48 experiments well. For the thermal dose, a suited $\sigma_d$ of 0.27 is identified (Figure 5.6 (a-c)). The corresponding value for the temperature $\sigma_t$ is 1.0. However, this width leads to values above 0 for a heat sink value of 1.0. Hence the function is scaled with an additional term which is 0.0 for $g(x) = 1$:

$$I^\sim_t(x) = I_t(x) e^{-\frac{g(x)^2}{2\sigma_t^2}} \sqrt{(1 - g(x))}$$

(5.5)

The resulting function $h_t$ is depicted in Figure 5.6 (d-f).

**Visualization:** In the fragment shader, the perfusion field is available as a 3D texture. Hence, for a fragment $x$, $g(x)$ is retrieved using a texture lookup function. The two transition functions $h_t$ and $h_d$ are sampled and stored into 1D-arrays, which are also provided to the shader as textures. Hence, the final energy value is computed as follows:

- Temperature:
  $$I^\sim_t(x) = I_t(x) h_t(g(x))$$
  (5.6)

- Thermal dose:
  $$I^\sim_d(x) = I_d(x) h_d(g(x))$$
  (5.7)

So far, the method has been described for one sonication. This approach is expanded to multiple sonications by concatenating respective shader code snippets in order to create one fragment shader which accumulates $I^\sim_t(x)$ or $I^\sim_d(x)$ accordingly. In the case of the temperature, the values of the individual sonications are combined using the max operator. For the thermal dose, the values are added and clamped above 1.0.

Finally, the resulting overall energy value for a fragment is mapped to a color using a transfer function.
Figure 5.6: Transition functions $h_d$ (a, c, e) and $h_t$ (b, d, f) for thermal dose and temperature, respectively. The scatter plots show the relation between energy values without perfusion ($T / Dose$), the perfusion values (heat sink) and the simulation values under perfusion influence (vertical axes, $T(heat\ sink) / Dose(heat\ sink)$).
5.4 Results

The method has been prototypically implemented using *MeVisLab* [164, 139]. A numerical simulation of HIFU sonications [4] as well as the perfusion field computation [12] already exist as MeVisLab components and were used accordingly. The user can position the transducer interactively by drawing a line starting from the transducer origin to the target region. After that, sonications can be placed with a click and dragged using the mouse.

Two lookup tables with varying resolutions have been computed for the 256 element phased array transducer by IGT / Imasonic using the above mentioned numerical simulation by Georgii et al. [4] for the computation of the temperature and thermal dose fields. The thermal dose is computed using the Arrhenius formulation [95]. A resolution of $256 \times 256 \times 256$ voxels has been used for the simulation. The energy profiles $e \in \mathbb{R}^{u \times v}$ have been sampled and stored with a resolution of $u \times v = 16 \times 4$ pixels.

*Lookup table 1* consists of 5488 samples and covers a relatively large region of the possible treatment area (80 \times 80 \times 100 \text{mm}, see Table 5.1, top four rows). *Lookup table 2* covers a smaller region (49 \times 49 \times 42 \text{mm}) which results in a higher spatial resolution (see Table 5.1, lower four rows). Furthermore, the distance between samples in the time dimension has been reduced from 20s to 10s compared to lookup table 1. As a result, lookup table 2 consists of 7056 samples. Figure 5.7 illustrates the domains covered by the two lookup tables in relation to the transducer. Furthermore, the setting used for the examples described in Section 5.4.2 is shown including the placement of the transducer with respect to the image data. The CT image used for the examples consists of 512 \times 512 \times 172 voxels with a voxel size of 0.625 \times 0.625 \times 1.0 \text{mm}. The used input segmentation masks are shown as red outlines in Figure 5.7 (the skin mask is not visualized).

<table>
<thead>
<tr>
<th>$P$</th>
<th>$t$</th>
<th>$d_w$</th>
<th>$d_a$</th>
<th>$p_x$</th>
<th>$p_y$</th>
<th>$p_z$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Min</td>
<td>150 W</td>
<td>20 s</td>
<td>25 mm</td>
<td>25 mm</td>
<td>−40 mm</td>
<td>−40 mm</td>
</tr>
<tr>
<td>Max</td>
<td>200 W</td>
<td>40 s</td>
<td>50 mm</td>
<td>50 mm</td>
<td>40 mm</td>
<td>40 mm</td>
</tr>
<tr>
<td>Samples</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>2</td>
<td>7</td>
<td>7</td>
</tr>
<tr>
<td>Distance</td>
<td>50 W</td>
<td>20 s</td>
<td>25 mm</td>
<td>25 mm</td>
<td>11.4 mm</td>
<td>11.4 mm</td>
</tr>
</tbody>
</table>

<table>
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<tr>
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<th>$t$</th>
<th>$d_w$</th>
<th>$d_a$</th>
<th>$p_x$</th>
<th>$p_y$</th>
<th>$p_z$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Min</td>
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<td>20 s</td>
<td>25 mm</td>
<td>25 mm</td>
<td>−24.5 mm</td>
<td>−24.5 mm</td>
</tr>
<tr>
<td>Max</td>
<td>200 W</td>
<td>40 s</td>
<td>50 mm</td>
<td>50 mm</td>
<td>24.5 mm</td>
<td>24.5 mm</td>
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<tr>
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<td>10 s</td>
<td>25 mm</td>
<td>25 mm</td>
<td>7 mm</td>
<td>7 mm</td>
</tr>
</tbody>
</table>

Table 5.1: Summary of lookup table 1 (*top four rows*) and lookup table 2 (*lower four rows*) for the IGT / Imasonic transducer.
Chapter 5. Interactive Visualization of High-Intensity Focused Ultrasound Sonifications

Figure 5.7: Illustration of the domains (green, dashed rectangle) covered by lookup table 1 (a) and lookup table 2 (b): the green dots show the spatial positions of the lookup table samples. The transducer is depicted by its origin and principal axis (green arrow) as well as the transducer elements that are partially contained in the displayed slice (circularly arranged gray rectangles).

5.4.1 Performance

On a computer with an Intel® Core™ i7 940 and a NVIDIA GeForce® GTX 670, the computation of the lookup tables 1 and 2 requires 12 and 17 hours, respectively. Hence, the simulation and analysis of a single sonication in the phantom is carried out in \(~8\) s. The reduction of the seven-dimensional lookup tables to the five-dimensional versions after transducer placement is completed in less than one second. However, the duration of the perfusion field computation is in the range of a few minutes (100 s for \(190 \times 160 \times 80\) voxels).

The interpolation and visualization during interactive sonication manipulation is carried out with \(~60\) frames per second. Even if 100 sonications are placed, the frame rate is still in the range of \(~30\) frames per second.

5.4.2 Examples

In the following several examples are presented. All presented figures arrange subfigures in three columns: the simulation results are shown in the left column, the approximation result in the center column, and a difference image in the right column. Temperature and thermal dose are displayed in the top and bottom rows,
respectively. The lookup tables used to apply colors to the values cover a range of $[37{}^\circ C, 100{}^\circ C]$ and $[0, 1]$ for the temperature and thermal dose, respectively. For the difference images, ranges of $[-20{}^\circ C, 20{}^\circ C]$ for the temperature differences and $[-1, 1]$ for the thermal dose differences are used. The difference images contain additional information about the maximum values of the corresponding simulation (“Max Sim”) and approximation (“Max Approx”) and the maximum signed differences (“Min Diff” and ”Max Diff”).

In Figure 5.8 an exemplary result for the approximation of a single sonication using lookup table 1 in comparison with the numerical simulation is depicted. Both methods consider a material image which labels adipose, liver, and vascular structures. The scenario is chosen such, that interpolation in all dimensions is necessary: the determined media layer depths are 30 mm and 32 mm for water and adipose, respectively. A sonication power of 180 W and a duration of 35 s is used. The sonication is placed between the spatial grid points. The size as well as the

![Figure 5.8](image)

Figure 5.8: Comparison of the numerical simulation (a, d) with the according approximation using lookup table 1 (b, e) for a single sonication (the focal spot is depicted by the small circle): the upper row shows the temperature, while the lower row shows the thermal dose. The rightmost column shows the difference between the simulation and approximation results.
Chapter 5. Interactive Visualization of High-Intensity Focused Ultrasound Sonications

shape, position, and orientation, as well as the temperature profile match closely. The result for the thermal dose (Figure 5.8 (e)) is inferior, however. The maximal thermal dose of the sonication is much lower than that computed by the numerical simulation (Figure 5.8 (d)). If the approximation is carried out using lookup table 2, the thermal dose is better represented (Figure 5.9 (e)). Although the thermal dose profile is still slightly different, the maximal thermal dose values match better. The crucial factor for the better results is the increased temporal resolution. As the thermal dose does not depend on time in a linear way [84], the linear interpolation in the time dimension cannot produce correct results for the thermal dose. However, the inaccuracies introduced by the interpolation can be reduced by increasing the sampling rate. Consequently, lookup table 2 is used for the remaining experiments.

Figure 5.10 shows the results for two sonications close to vascular structures. It can be seen, that the heat sink effect is considered by the approximation. The agreement between the results of the approximation and simulation are good with respect to the temperature estimation. The thermal dose estimation exhibits similar outlines, but the values inside are different. In general, the approximation underestimates the

![Figure 5.9](image)

Figure 5.9: Results generated using lookup table 2 for the same sonication as depicted in Figure 5.8. Consequently, the results of the numerical simulation (a, d) are identical to those in Figure 5.8.

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Figure 5.10: Representation of the heat sink effect for two sonications (sonication 1: temperature in row 1, thermal dose in row 2, sonication 2: temperature in row 3, thermal dose in row 4).
thermal dose values. This effect is increased in the proximity of vascular structures. The combination of multiple sonications is depicted in Figure 5.11. Here, differences between approximation and simulation are visible for both, temperature and thermal dose. The deviations regarding thermal dose are a direct consequence of the observed inaccuracies. However, the strong difference in the temperature results from the temporal course of the experiment. The numerical simulation considers cooling after a sonication is finished. Hence, figure (a) mainly shows the heat resulting from the last of the planned sonications, while the approximation can only display the maximum temperature over time.

Another restriction of the approximation method is visible in Figure 5.11: additional, unplanned regions of strong heating, which may result from strong absorption, or reflection, are predicted by the numerical simulation, but not by the approximation. As can bee seen in the images (a) and (c), the placement of the sonications close to the liver capsule leads to heating at the interface, which, accumulated over the course of six sonications, also amounts to ~60°. This effect is not considered by the approximation at all.

![Figure 5.11: Simulation (a, d) and approximation (b, e) of a plan consisting of three sonications.](image)

(a) ![Simulation](image)
(b) ![Approximation](image)
(c) ![Dose](image)
(d) ![Simulation](image)
(e) ![Approximation](image)
(f) ![Dose](image)
5.5 Discussion and Conclusion

In this chapter, a method for the interactive approximation of the results of a numerical simulation for the planning of HIFU treatments is presented. The method has been implemented exemplary for the liver. However, a translation to other treatment scenarios should be possible. Most likely, this would involve the adaption of the media layer model, which, at the moment, only considers two tissue layers. In principle, there is no restriction regarding the number of layers. Since the corresponding dimensions can be reduced once the transducer has been placed, there should be no noticeable impact on the performance of the method. Of course, more layers would increase the preprocessing time. However, the preprocessing has only to be carried out once per transducer. The experiments carried out in this work do not consider the material of the tumor, which is not identical to the liver. An according extension would also be possible using the layer model. This model has one serious constraint though: it assumes a fixed order of layers. Hence, concave parts of organs, or situations in which the order of layers might vary, are not supported yet.

Numerical simulation requires at least several seconds for the computation of a single sonication, even using highly parallelized implementations and current GPUs. The proposed method allows for an interactive preview and may therefore complement numerical approaches. It facilitates the modification of individual sonications at very high frame rates. The impact of the number of sonications is low because only the interpolation of the changed sonication has to be updated. Only the performance of the rendering using the modified fragment shader slightly depends on the number of displayed sonications, since more computations have to be carried out per fragment. However, even for very large numbers of sonications, the frame rate is still high and allows for real time feedback.

The method is still subject to certain inaccuracies. First, there are principal restrictions resulting from the general approach. The proposed method only tries to predict the planned sonication. Reflection and absorption might however lead to heating outside the target region and might reduce the energy deposited in the target region. To evaluate such effects, a numerical simulation is still needed. Hence, the proposed method aims at supporting the simulation based planning workflow, not at replacing it.

However, in addition to these fundamental restrictions, the method suffers from inaccuracies that are a result of the current implementation, not of the general approach itself. One source of inaccuracies is the relatively low resolution of the stored energy profiles ($16 \times 4$ pixels). The interpolation carried out by the texture lookup results in a smooth appearance, but cannot accurately reconstruct the original energy distribution. Hence, the utilization of higher resolutions should be investigated. Especially the estimation of the thermal dose can be further improved. It should be possible to improve the overall results by using an even higher resolution in the time dimension. The currently used transition function for the modeling of the influence of the perfusion on the thermal dose field is a fixed function that does not depend
on the material layers or the sonication settings. It should be investigated, if the usage of a multi-dimensional lookup table can improve the results.

Contributions and Publications An early proof of concept implementation of the proposed method has been presented at the Society for Thermal Medicine’s 31st Annual Meeting (STM) in 2014 [20]:


This implementation only considered the interpolation of ellipsoids that correspond to completely ablated regions. Energy profiles, heterogeneous material or heat sink effects were not considered. The method was partially implemented by Eicke Mücke, a student which was supervised by the PhD candidate. The method presented here is currently prepared for submission to the International Journal of Hyperthermia.
6 Conclusion

All models are approximations. Essentially, all models are wrong, but some are useful. However, the approximate nature of the model must always be borne in mind.

(George E.P. Box)

Tissue ablation procedures allow to destroy pathological tissue deep in the body by only using very thin needles, or even without harming the skin at all. What would have been described as magic or science fiction a hundred years ago, is an important part of the clinical practice today. With the fantastic possibilities also come challenges: The planning and implementation of such procedures in a manner that destroys all pathological tissue in the target volume and spares all non-target structures requires both time and expertise. The aim of this PhD thesis is to provide methods that facilitate the planning of successful and safe ablation procedures. To this end, visualization and optimization approaches for the selection of suited access paths for needle-based interventions as well as for HIFU sonication planning are presented.


text

6.1 Challenges

Planning is an important step of any ablation procedure. As the anatomical situation is not directly visible during the procedure, an exact strategy has to be laid out beforehand based on preprocedural image data. An access route to the target has to be identified which spares non-target structures with respect to penetration by the instrument or unwanted heating by the HIFU beam. The position and extent of the probable coagulation resulting from the therapy has to be predicted and adapted accordingly to facilitate complete ablation of the target volume. If a single ablation zone is smaller than the target volume, this process requires the composition of multiple ablation zones.

Hence, this thesis focuses on computer-assisted planning of tissue ablation procedures. This area of research comprises many aspects, and only a subset is considered in this PhD thesis. The following clinical problems are addressed:

- Identification of suited, straight access paths for ablation procedures that use needle-like instruments. Various criteria have to be considered during this process.

- Determination of sonication plans during planning of HIFU procedures. This requires the combination of many small sonifications in order to ablate the whole target volume.

In order to enhance the clinical planning workflow without disturbing it the developed methods should fulfill the following requirements:

- The computational complexity must be kept low in order to meet the restricted time window available for planning.

- The methods should be easy to use. Complex parametrization of algorithms should be avoided.

- The methods must not lead to decisions impairing the safety of the patient. Ideally, they should help to make the procedure safer.

6.2 Contributions of this Thesis

This PhD thesis investigates interactive visualization methods and heuristic optimization approaches for the support of physicians during planning of minimally- and non-invasive tissue ablation procedures. The first method developed in the context of this thesis facilitates a guided manual access path selection process. It introduces the risk structure map concept, which allows to highlight infeasible skin entry points directly in the established 2D image viewers. Based on a user-defined target point, the method uses volume rendering and the cube mapping approach to project the risk structures onto the currently
displayed 2D image slice. Any point covered by the projection of a risk structure is displayed red indicating that this point is not suited as a skin entry point since the resulting path would penetrate a risk structure. The visualization is carried out in real time and does not require any complex preprocessing steps. An evaluation study shows, that this method is suited to potentially reduce planning times while raising the safety of the determined paths. This method is further advanced into a comprehensive visualization method that also supports the definition of the target point. The complete geometry of the instrument is considered. Hence, all possible collisions are detected, including those with other instruments. Therefore, the planning of ablations of larger tumors, which requires the placement of multiple instruments, is facilitated.

The underlying idea of the risk map also forms the basis of an optimization approach which uses GPU-accelerated projections and simple image processing methods. This fast approach allows to compute suited access paths under consideration of multiple clinical criteria in a few seconds. A simple intuitive user interface allows to iterate over a list of maximal three access paths. Two evaluation studies show that the method reliably computes proposals that are comparable to access paths as defined by radiologists with the experience of several years. A second optimization approach is presented which aims at the integration of an ablation zone prediction method that is based on simulation. Furthermore, the incorporation of interactive methods for the exploration of the solution space is investigated. To this end, the first projection-based approach is combined with a numerical optimization scheme which allows to approximate the Pareto front. A study showed, that the interactive exploration is appropriate to select adequate access paths. However, the reproducibility of the interaction as well as the computation time show potential for improvement. Overall, the quality of the results is inferior compared to those of the first, projection-based method.

For the assistance of sonication planning in the context of HIFU procedures, an approximation of the numerical simulation of temperature and thermal dose is proposed. It is based on the analysis of the actual numerical simulation of a limited number of example configurations. This time consuming process is carried out once for each transducer type and results in large lookup tables. During interactive sonication planning, these lookup tables are used to reconstruct the temperature or thermal dose field for a certain sonication. This field is combined with a perfusion representation to incorporate heat sink effects based on a patient individual vessel mask. The method considers tissue layers along the beam path, but does not incorporate effects of structures that lead to unwanted absorption or reflection such as bones or the bowel.

\section{Discussion}

All methods presented in this thesis are developed with a focus on applications in the liver. However, only few details are really liver specific. Hence, a translation to
other treatment scenarios is feasible. Furthermore, all approaches for the definition of access paths were developed in order to support RFA. The reasons for that have a historical background: RFA is the most popular ablation modality and dominated the according research projects. However, from a technical point of view the proposed methods can be applied to most other needle-based ablation procedures. Both, the advanced risk structure map method as well as the projection-based optimization are integrated into the intervention planning platform SAFIR, which generally supports all of the needle-based modalities which have been discussed in Section 2.1.4.

Another similarity of all proposed methods is the nature of the required inputs: considered anatomical structures have to be provided by means of segmentation masks. The creation of these masks is not in the scope of this work. In general the achievable quality of the results of the presented methods depends on the quality of the input segmentation masks. Although some of the methods exhibit a high robustness with regard to smaller segmentation deficits, the usage of accurate input masks is mandatory in order to use the presented methods to their full potential. It could be argued that the time required to compute or manually define the segmentation masks diminishes the benefit of using the proposed methods. However, fully automatic segmentation tasks can be carried out in the background. During this period, the radiologist can already examine the image data and measure or segment the tumor.

Two assumptions are central to the access path planning methods presented here: The shaft of the instruments can be modeled as straight, rigid cylindrical objects. Furthermore, the anatomy as depicted in the preprocedural images corresponds to the anatomical situation encountered during the procedure. However, both the instrument as well as the patient anatomy are subject to deformations [30]. Recently, a few works try to model these deformations [92, 126]. However, none of these works discusses how the needed material parameters can be determined in a patient specific manner. Furthermore, the modeling of the interaction between the organs and tissues is in an early stage. While the investigation of ways to incorporate deformations during access path determination is an important field of research, the methods presented here follow a different route that has also been adapted by most other work in this field: support the identification of a linear path that allows for deviations by favoring larger distances to risk structures. Furthermore, all of the methods presented here aim at planning based on images acquired shortly before the procedure. If the same state of inhalation is used during planning and intervention, the risk of anatomy changes is further minimized.

### 6.4 Outlook

As discussed before, methods that support the planning of ablation procedures, typically depend on segmentation masks. The acceptance of such planning approaches in the clinical routine will depend on the availability of fully automatic, or fast and easy to use manual and semi-automatic segmentation tools. For example, for hepatic interventions the segmentation of the whole ribcage is crucial. While bony structures
can be separated from the background, the segmentation of the cartilage is very challenging. Manual segmentation is not really an alternative here because of the complex anatomy and the bad visibility in CT-scans. Currently automatic approaches to that challenging problem are investigated at Fraunhofer MEVIS. Likewise, the translation of planning methods to other anatomical sites will also require advances in the area of segmentation.

In the case of needle-based interventions, the treatment plan has to be manually implemented, typically. As discussed before, the translation of planning results into the intervention is subject to several challenges. In an ideal scenario, a defined plan can be directly implemented. However, even in this case, the translation of the plan might be difficult since the planned point of incision has to be identified on the patient skin and the planned angulation has to be achieved by moving the instrument handle accordingly. Hence, the development of methods that improve the transfer of the plan into the interventional scenario is a big challenge. To mitigate the consequences of deformations, the intraprocedural adaption of treatment plans should be investigated. Furthermore, the knowledge about potential deformations might be integrated into existing planning methods in the form of non-uniform safety margins.
Appendix
## List of Abbreviations

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<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
<th>Page</th>
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<tbody>
<tr>
<td>API</td>
<td>Application programming interface</td>
<td>27</td>
</tr>
<tr>
<td>CPU</td>
<td>Central processing unit</td>
<td>80</td>
</tr>
<tr>
<td>CT</td>
<td>Computed tomography</td>
<td>8</td>
</tr>
<tr>
<td>DTF</td>
<td>Distance transform</td>
<td>50</td>
</tr>
<tr>
<td>DVR</td>
<td>Direct volume rendering</td>
<td>26</td>
</tr>
<tr>
<td>FEM</td>
<td>Finite element method</td>
<td>74</td>
</tr>
<tr>
<td>GLSL</td>
<td>OpenGL Shading Language</td>
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</tr>
<tr>
<td>GPU</td>
<td>Graphics processing unit</td>
<td>4</td>
</tr>
<tr>
<td>GUI</td>
<td>Graphical user interface</td>
<td>5</td>
</tr>
<tr>
<td>HIFU</td>
<td>High-intensity focused ultrasound</td>
<td>2</td>
</tr>
<tr>
<td>IRE</td>
<td>Irreversible electroporation</td>
<td>10</td>
</tr>
<tr>
<td>LITT</td>
<td>Laser interstitial thermal therapy</td>
<td>15</td>
</tr>
<tr>
<td>MinIP</td>
<td>Minimum intensity projection</td>
<td>26</td>
</tr>
<tr>
<td>MIP</td>
<td>Maximum intensity projection</td>
<td>26</td>
</tr>
<tr>
<td>MOO</td>
<td>Multi-objective optimization</td>
<td>30</td>
</tr>
<tr>
<td>MPR</td>
<td>Multi planar reformation</td>
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<tr>
<td>MR</td>
<td>Magnetic resonance</td>
<td>25</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic resonance imaging</td>
<td>8</td>
</tr>
<tr>
<td>MWA</td>
<td>Microwave ablation</td>
<td>2</td>
</tr>
<tr>
<td>PACS</td>
<td>Picture archiving and communication system</td>
<td>23</td>
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<tr>
<td>PET</td>
<td>Positron emission tomography</td>
<td>19</td>
</tr>
<tr>
<td>RFA</td>
<td>Radiofrequency ablation</td>
<td>2</td>
</tr>
<tr>
<td>SOO</td>
<td>Single-objective optimization</td>
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</tr>
<tr>
<td>SPECT</td>
<td>Single-photon emission computed tomography</td>
<td>19</td>
</tr>
<tr>
<td>US</td>
<td>Ultrasound</td>
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</tr>
<tr>
<td>VOI</td>
<td>Volume of interest</td>
<td>122</td>
</tr>
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List of Own Publications


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[166] Rossi, S., Fornari, F., Pathies, C., and Buscarini, L. “Thermal Lesions Induced by 480 KHz Localized Current Field in Guinea Pig and Pig Liver.” In: *Tumori*, 76(1) (Feb. 1990), pp. 54–57 (cited on p. 8).


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