Doctoral Thesis

Scientific bases for dosimetric compliance tests of mobile telecommunications equipment

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SCIENTIFIC BASES FOR DOSIMETRIC
COMPLIANCE TESTS OF MOBILE
TELECOMMUNICATIONS EQUIPMENT

A dissertation submitted to the
SWISS FEDERAL INSTITUTE OF TECHNOLOGY
ZURICH

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Doctor of Technical Sciences

presented by
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1996
# Contents

Summary v  

Zusammenfassung vii  

Acknowledgments ix  

I Introduction 1  

1 Background and Motivation 3  

1.1 Introduction .................................................. 4  

1.1.1 Mobile Telecommunications Equipment ..................... 4  

1.1.2 Electromagnetic Energy Absorption, Limits and Exclusions .. 6  

1.1.3 Nearfield Absorption .................................... 8  

1.1.4 Preliminary Tests of Modern MTEs ......................... 9  

1.2 Compliance Tests ........................................ 9  

1.2.1 Basic Requirements for Standardization .................. 9  

1.2.2 Difficulties in Determining the SAR ..................... 11  

1.2.3 CENELEC/WGMTE Proposal ............................... 13  

1.2.4 Conclusion .............................................. 13  

II Increase in Measurement Accuracy 19  

2 Broadband Calibration of E-Field Probes in Lossy Media 21  

2.1 Introduction ............................................. 21  

2.2 Calibration Requirements .................................. 22  

2.3 Three-Step Calibration Approach ............................ 24  

2.4 Calibration in Air ......................................... 26  

2.5 Experimental Determination of Conversion Factor $\gamma$ ........ 26  

2.5.1 Setup .................................................. 26  

2.5.2 Brain-Simulating Liquids ................................. 28  

2.5.3 Temperature Probe ..................................... 29  

2.5.4 Results ............................................... 30  

2.6 Numerical Determination of Conversion Factor $\gamma$ ........ 31  

2.6.1 Modeling of the Probe .................................. 32  

2.6.2 Results of the Simulations ............................... 34  

2.7 Conclusions ............................................... 36
III Head Modeling

3 The Dependence of EM Energy Absorption upon Human Head Modeling at 900 MHz

3.1 Introduction .............................................. 43
3.2 Numerical and Experimental Techniques .................. 45
  3.2.1 Simulation Technique ............................... 45
  3.2.2 Experimental Technique ............................. 47
3.3 Head Phantoms ........................................... 48
  3.3.1 Exposure ........................................... 48
  3.3.2 Numerical Phantoms ................................. 48
  3.3.3 Experimental Phantoms .............................. 51
3.4 Results .................................................. 53
  3.4.1 SAR Distribution in Various Head Phantoms .......... 53
  3.4.2 Averaged SAR Values ................................ 54
3.5 Conclusion ............................................... 58

4 The Dependence of EM Energy Absorption upon Human Head Modeling at 1800 MHz

4.1 Introduction .............................................. 61
4.2 Methods .................................................. 62
  4.2.1 Numerical and Experimental Techniques ............. 62
  4.2.2 Head Phantoms ..................................... 62
4.3 Results and Discussion .................................. 65
4.4 Conclusions .............................................. 68

IV Homogeneous Modeling For Compliance Tests

5 Consideration of Additional Factors

5.1 Introduction .............................................. 75
5.2 Variations in the Ear Region ................................ 76
  5.2.1 SAR-Estimation in the Ear Region .................... 77
  5.2.2 Conclusions for Compliance Tests ................... 78
5.3 Absorption in Children .................................. 79
  5.3.1 Child Brain Tissue Parameters ....................... 81
  5.3.2 Child Head Geometry ................................ 81
  5.3.3 Conclusions for Compliance Tests ................... 82
5.4 Additional Non-Anatomical Factors ......................... 82
  5.4.1 The Hand Holding the Device ......................... 83
  5.4.2 External Metallic Structures ......................... 84
  5.4.3 Internal Metallic Structures ......................... 88
5.5 Conclusion ............................................... 90

6 Uncertainties and Critical Design Factors

6.1 Estimation of Uncertainty ................................ 93
  6.1.1 SAR Assessment Method .............................. 94
  6.1.2 Anatomical Simplification ........................... 94
  6.1.3 External/Internal Structures ....................... 95
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Abstract

The massive growth in the use of handheld mobile telecommunications equipment (MTE) has subjected the general public to drastically increased electromagnetic exposure. In view of this, public health organizations began to call for compliance tests of MTEs as the most sound approach to satisfy both consumer safety and industry interests.

A dosimetric measurement system for compliance testing was developed between 1992 and 1994 based on the conclusions of a previous study on nearfield absorption at frequencies above 300 MHz. Not yet determined was the question of the standardized modeling of the absorption in the human body which is addressed in this thesis. The goal was to set the scientific basis for compliance testing which would meet the requirements defined by standards organizations. The difficulties in this project arose from minimizing overestimations in the induced specific absorption rate (SAR) of the simulated exposition while, at the same time, covering all users under all operational conditions. This problem must be understood together with the fact that the absorptions caused by modern mobile telecommunications equipment actually come very close to today’s safety limits. To meet the envisioned goal several detailed studies were necessary:

The first part involved analyzing the calibration procedures of E-field probes in air and simulating liquids leading to the development of a new improved calibration procedure. This yielded a considerable improvement (by a factor of two) of the calibration uncertainty of previously known methods and led to an increase of the SAR assessment accuracy within homogeneous shell phantoms.

The second part investigated the influence of head modeling on the induced SAR. In particular the possibility of assessing the maximum exposure in the human head with simple homogeneous phantoms and the amount of uncertainty involved were determined. Two different numerical simulation tools were used (MAFIA and MMP) and the dosimetric scanner DASY2 was employed as an experimental tool. To gain model-independent results, four numerical head models, three experimental head models and various layered spheres were available.

The third part addressed whether homogeneous models based on the results of the previous investigations enable the coverage of additional influences on the induced SAR in the head: the hand of the user holding the device, small parallel shifts of the device in close proximity to highly anatomically heterogeneous head regions, influence of external metallic structures, such as glasses and jewelry, and internal structures, such as metallic implants.

The results can be summarized as follows: Homogeneous shell phantoms are suitable for compliance tests of MTEs. The maximum exposure in all users can be reliably assessed with small uncertainties. Furthermore, with respect to compliance tests homogeneous models are advantageous in comparison to heterogeneous models:
1) they are insensitive to small parallel shifts of the device. Thus, the number of measurements necessary to assess the maximum SAR value can be considerably reduced. 2) Increases in absorption due to external metallic objects (glasses, jewelry) and internal objects in the mouth and chin region are automatically covered when using homogeneous models. 3) It is not necessary to simulate the hand holding the device. It has been shown that the worst case in respect to the power absorbed in the head where a hand is involved is similar to the case with no hand involved. 5) Only internal metallic structures in the upper head region can not be covered by homogeneous models and are therefore excluded from this issue.

The standardization process has lasted longer than expected and a final version has not yet been agreed upon. Nevertheless, all relevant issues for compliance tests have been addressed and worked out in detail in this project. The scientific and technical basis for standardized compliance testing of MTEs is now at hand.
Zusammenfassung

Die massive Zunahme im Gebrauch von Mobiltelefonen bedeutet für die allgemeine Bevölkerung eine drastisch erhöhte elektromagnetische Belastung. Entsprechend begannen verschiedene Gesundheitsbehörden eine Typenprüfung für mobile Kommunikationssysteme wie Mobiltelefone (Handy) zu fordern. Dies wurde als der vernünftigste Ansatz im Sinne des Konsumentenschutzes wie auch im Interesse der Industrie angesehen.


Im ersten Teil wurden die bisher bekannten Kalibrierungsverfahren für E-Feld Sonden in Luft und in Flüssigkeiten um einen Faktor zwei entscheidend verbessert. Diese Verbesserung führte zu einer erhöhten Genauigkeit der Bestimmung der Absorptionsverteilung in homogenen Schalenmodellen.

Im zweiten Teil wurde der Einfluss der Kopfmodellierung auf die induzierte spezifische Absorptionsrate bestimmt. Insbesondere wurde untersucht, ob die höchste Belastung mit stark vereinfachten Kopfmodellen, d.h. mit homogenen Phantomen, zuverlässig bestimmt werden kann und falls ja, mit welcher Unsicherheit. Dazu wurden zwei voneinander unabhängige numerische Simulationsverfahren (MAFIA und MMP) sowie der dosimetrische Scanner DASY2 eingesetzt. Um möglichst modellunabhängige Resultate zu erhalten, wurde mit vier unterschiedlichen numerischen Kopfmodellen, drei experimentellen Kopfmodellen plus verschiedenen Kugelschalenmodellen gearbeitet.

Im dritten Teil wurde basierend auf den vorausgegangenen Resultaten untersucht, ob homogene Modelle in der Lage sind zusätzliche Faktoren, die die im Kopf induzierte SAR beeinflussen, zu erfassen: der Einfluss der Hand, die das Gerät hält, das Erfassen von kleinen Parallelverschiebungen der Quelle in Kopfregionen mit sehr komplexer heterogener Gewebeverteilung, der maximale Einfluss externer metallischer Objekte (z.B. Brillen und Schmuck) und der Einfluss von Implantaten.
Die Resultate lassen sich wie folgt zusammenfassen: Homogene Schalenphantome sind für die Typenprüfung geeignet, d.h., die höchste Belastung unter allen Benutzern kann zuverlässig mit kleiner Unsicherheit im Vergleich zum real auftretenden Wert bestimmt werden. Zudem zeigen homogene Modelle verglichen mit heterogenen Modellen im Hinblick auf Typenprüfungen Vorteile: 1) Sie sind gegenüber kleinen Parallelverschiebungen der Geräte am Kopf unempfindlich. Damit wird die Anzahl notwendiger Messungen erheblich reduziert. 3) Die Überhöhungen der Absorption durch externe metallische Objekte (Brille, Schmuck, etc.) und im Kieferbereich implantierte metallische Strukturen werden mit homogenen Phantomen automatisch berücksichtigt. 4) Da die ungünstigste Art, das Gerät zu halten, jener Exposition gleichkommt, wie wenn das Gerät nicht von der Hand berührt wird, ist es von Vorteil, die Hand nicht zu modellieren. 5) Aus den Untersuchungen über metallische Implantate im Ohr- und Hirnbereich muss gefolgt werden, dass dieses Problem nur mit extremem Aufwand berücksichtigt werden könnte, und daher ausgeschlossen werden muss.

Damit sind die wissenschaftlichen und technischen Grundlagen und Voraussetzungen geschaffen und einer Einführung einer Typenprüfung steht von dieser Seite nichts mehr im Wege.
Acknowledgments

This work would not have been possible without the help of many people involved throughout the entire project. First I would like to mention "the team" to whom I express my deep gratitude. Thomas Schmid, the engineering brain of the group, the man behind the novel E-field probe and the dosimetric scanner DASY. His knowledge and experience provided both essential input and a source of solutions during the whole process. I would like to thank Dr. Ralf Kästle, who joined the team 12 months ago, for his help and his down-to-earth personality; Michael Burkhardt whose expertise in the MAFIA code was invaluable for this work; Roger Tay from Motorola Inc. receives my warm thanks as the guy on the MMP side; Hansruedi Benedickter the “good spirit from upstairs” for his measurement support; Jeroen de Keijzer for his various constructions; Walter Nusser for the experimental head models; Oliver Egger, the creator of DASY’s software, for his programming talents and his attempt to overcome the robot’s noise in the lab by cool music; Katja Poković for committing so many long hours to the lab; Thomas Schwitter for the Mac support, the layout of reports and papers and, as the New Yorkers might say, for being “such a Mensch”. Without this team mobile phones would still heat brains. And, of course to make a fabulous team complete, the team leader, Prof. Niels Kuster. I would like to express my sincere gratitude to him for guiding me through this work. I was fortunate to be able to profit from his great scientific competence and his introduction to the challenging balancing act independent academics must master when investigating areas that touch on both industry interests and consumer safety. His sense for such political arenas allowed me to fully appreciate the motivating field that lay beyond the laboratory environment where public interest was always inherent to my work.

I would like to express my appreciation to Prof. Heinrich Baggenstos, head of the institute, for providing us with the necessary freedom in the early days of these studies. I would like to thank Ray Ballisti for system administration and all other members of the institute for the amicable atmosphere.

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Taylor and Dr. Horst Simon for their advice and for companionship and friendship
which were kept intact all through these years even when the pint turned virtual.
Part I

Introduction
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Chapter 1

Background and Motivation

Abstract- In recent years handheld mobile telecommunication equipment (HMTE) has become a common and widely used consumer product. Falling operational costs will lead to more frequent and more prolonged conversations. However, the operation of transmitters in very close proximity to the head subjects the user to an electromagnetic exposure with induced field strengths that are significantly higher than any to which the general public has been exposed in an “everyday” environment before. The maximum power absorbed by the user is limited by internationally agreed safety limits. Nevertheless, recent studies demonstrated that modern telecommunication devices can violate those limits. In view of this, some health agencies called for dosimetric type approval and require compliance with safety limits under all operational conditions. This approach was fostered by several law suits against manufacturers of devices and providers of mobile communication networks which sent a shockwave through the unprepared telecommunications industry by associating the use of mobile phones to cancer (1992). Though the cancer question is not being addressed in the discussion of current safety limits, it reflects the economic and political sensitivity of this topic. Based on earlier studies on the absorption mechanism in the close nearfield, a dosimetric scanner was developed between 1992 to 1995 which enables a reliable, accurate and fast assessment of the absorption of electromagnetic energy in liquids. This dosimetric scanner is in use today for dosimetric tests in various industry laboratories in Europe and the United States and could be used for compliance tests. A simple homogeneous model is used to simulate the human body. This approach was regularly questioned regarding its suitability and accuracy in determining the actual exposure of a mobile telephone user. The following chapter gives an introduction to the issue of compliance tests, describes the initial situation and addresses the motivation behind this study.
1.1 Introduction

1.1.1 Mobile Telecommunications Equipment

Mobile communication has come a long way from simplex mobile radio communications for military, police, maintenance or dispatch purposes in the late thirties to today's handheld mobile communication devices marketed to the general public. A big technological as well as economic leap was taken when shifting from cellular car bound or portable devices to handheld mobile phones. Progress in the microwave semiconductor technology and in spread spectrum technology enabled the further miniaturization of handheld devices as well as increase in the number of users per bandwidth, thus decreasing access costs for the user at the same time. This progress, largely taking place within one decade, initiated a shift of mobile communications from a limited professional environment into the wide consumer world. The market penetration of mobile phones has reached levels of more than 20% as in the case of Sweden. In most countries the market growth figures are regularly upward-corrected to adjust to the high demand in modern mobile communications equipment (MTE). The growth rate in the United States has run between 20% and 30% in the last decade. The use of modern MTE is not limited to the developed world but shows even stronger growth in the developing countries where telecommunications policy is going to bypass the "tethered" telephone network by directly setting up wireless communication networks.

In 1995 about 53 million users subscribed to analog systems and about 15 million to digital systems worldwide. For the year 2000 a global forecast of mobile phone subscribers expects 65 million for analog systems and 50 million for digital systems.

Digital systems are taking the lead over analog systems. A prominent reason is their more efficient use of the limited bandwidth. Multiple access schemes such as TDMA (Time Division Multiple Access) and CDMA (Code Division Multiple Access) allow several subscribers to use the same single band and, therefore, increase the bandwidth efficiency. TDMA has been adopted for the originally pan-European GSM system (Global System of Mobile Communication). CDMA, as a more recent technology, will become a competitive digital system in particular in the United States. In addition to terrestrial mobile communications networks, satellite-based networks are currently being developed and will be put into service in the near future. Those systems use a network of satellites in a low earth orbit (LEOS) to link a mobile call from a handheld mobile phone to other mobile users (phone, pager, cars, aircrafts) or to connect with terrestrial gateways which, in turn, link them into the ordinary telephone network. Though the use of other wireless communication devices (pagers, wireless local area networks) go hand in hand with the above-mentioned systems, such systems do not fall within the scope of this study which addresses exclusively mobile communication systems which radiate in close proximity of the human head.

A wide variety of cellular systems is currently available worldwide (see Tables 1.1, 1.2). In Europe interoperability has become a dominant feature of modern cellular systems. Interoperability enables roaming in various countries. The same trend can be seen in Southeast Asia and East Asia. Thus GSM based systems were chosen as a digital enhancement of the existing national analog systems. In the USA capacity is of greater importance than interoperability. To achieve this goal dual mode AMPS-systems are planned which allow roaming between the analog and the digital portions
### Table 1.1: Current analog mobile communications systems. The frequency refers to the mobile. The power is the peak power of the handheld mobile.

<table>
<thead>
<tr>
<th></th>
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</tr>
</thead>
<tbody>
<tr>
<td>AMPS</td>
<td>USA</td>
<td>824-849</td>
<td>FDMA</td>
<td>0.60</td>
</tr>
<tr>
<td>E-AMPS</td>
<td>USA &amp; Southeast Asia</td>
<td>869-894</td>
<td>FDMA</td>
<td>0.60</td>
</tr>
<tr>
<td>TACS</td>
<td>Europe, Asia (China), Pacific, Middle East</td>
<td>872-905</td>
<td>FDMA</td>
<td>0.60</td>
</tr>
<tr>
<td>ETACS</td>
<td>France</td>
<td>890-915</td>
<td>FDMA</td>
<td>10</td>
</tr>
<tr>
<td>NMT450</td>
<td>Europe</td>
<td>451.5-443.55</td>
<td>FDMA</td>
<td>1</td>
</tr>
<tr>
<td>NMT900</td>
<td>Europe, Asia</td>
<td>890-915</td>
<td>FDMA</td>
<td>1</td>
</tr>
<tr>
<td>C-NET</td>
<td>Germany</td>
<td>451.3-455.74</td>
<td>FDMA</td>
<td>0.75</td>
</tr>
<tr>
<td>JTAC</td>
<td>Japan</td>
<td>898-925</td>
<td>FDMA</td>
<td></td>
</tr>
<tr>
<td>NTAC</td>
<td>Japan</td>
<td>898-925</td>
<td>FDMA</td>
<td></td>
</tr>
</tbody>
</table>

### Table 1.2: Current and future digital mobile communications systems. The frequency refers to the band used by the mobile. Chn/Car stands for Channels per Carrier. $P_{ave}$ is the time averaged maximum power. (n.e.y. = not established yet)

<table>
<thead>
<tr>
<th>System</th>
<th>Country</th>
<th>Frequency [MHz]</th>
<th>Multiple Access</th>
<th>Chn/Car</th>
<th>$P_{ave}$ [W]</th>
</tr>
</thead>
<tbody>
<tr>
<td>N-AMPS</td>
<td>The Americas</td>
<td>900</td>
<td>n.e.y.</td>
<td>n.e.y.</td>
<td>n.e.y.</td>
</tr>
<tr>
<td>D-AMPS</td>
<td>North America, Asia</td>
<td>900</td>
<td>n.e.y.</td>
<td>n.e.y.</td>
<td>n.e.y.</td>
</tr>
<tr>
<td>GSM</td>
<td>Worldwide</td>
<td>890-915</td>
<td>TDMA</td>
<td>8</td>
<td>0.250</td>
</tr>
<tr>
<td>DCS-1800</td>
<td>Europe</td>
<td>1710-1785</td>
<td>TDMA</td>
<td>8</td>
<td>0.125</td>
</tr>
<tr>
<td>PCS-1900</td>
<td>USA</td>
<td>1850-1910</td>
<td>TDMA</td>
<td>n.e.y.</td>
<td>n.e.y.</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1930-1990</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>IS-95</td>
<td>USA</td>
<td>869-894</td>
<td>CDMA</td>
<td>-</td>
<td>0.6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>824-849</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PDC</td>
<td>Japan</td>
<td>940-956</td>
<td>TDMA</td>
<td>3</td>
<td>0.66</td>
</tr>
<tr>
<td></td>
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<td>1453-1465</td>
<td></td>
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</tr>
</tbody>
</table>

Modern MTEs of a cellular network operate in the frequency range between 400 MHz and 2 GHz. Their time-averaged input power ranges from about 100 mW to a maximum 10 W (see Tables 1.1, 1.2). Cordless telephones (e.g. CT- and DECT systems) do not fall within the safety issue since their operational range is limited to a few hundred meters and thus their input power is considerably lowered.

The trend is tending towards worldwide accessibility through interoperability or satellite based systems for highly mobile professional users, and personal phone users who want high capacity and high mobility within a small operational range. The former will be mainly achieved by introducing global standards such as GSM and the latter by reducing cell sizes and increasing cell sites in highly populated areas, and shifting to higher frequency bands. Smaller cell cites in turn mean that the averaged input power of the handheld devices will further decrease.

The expansion of mobile communications equipment beyond mainly business to personal usage, together with falling operational costs, leads to more frequent and
more prolonged conversations. The operation of transmitters in very close proximity to the head causes an electromagnetic exposure of the user which is several magnitudes above the average background exposure. The locally induced field strength is significantly higher than any exposure the general public has been exposed to before, except through certain medical treatments (e.g., diathermy, hyperthermia [1] MRI [2]). Recent studies have even shown that some current cellular phones are close to or even exceed current limits under certain operational conditions [3].

1.1.2 Electromagnetic Energy Absorption, Limits and Exclusions

To protect the public from extensive exposure to electromagnetic fields the maximum tolerable exposure is limited. Up until 1982 the limits were simply defined by exposure quantities such as frequency, power densities and field strengths. In 1982 the American National Standards Institute (ANSI) introduced a dosimetric approach for non-ionizing radiation referring to the same concept as in ionizing radiation protection [4]. The dose is defined as the energy absorbed per unit mass. As it is the case in ionizing radiation, a direct correlation between dose and biological damage is assumed. In the frequency range of mobile communications biological damage is predominantly correlated with the temperature increase in tissue due to absorption of the incident electromagnetic energy. Biological tissue is primarily water containing (70 - 75%). The penetration depth of the absorbed energy in the mobile communications frequencies range from 20 mm to 40 mm. Since human temperature sensors are adjusted to infrared detection they are located in the outer skin layers. Therefore, the body is not “equipped” to react adequately to energy deposition in the frequency range of mobile communications.

A difficulty which arises for the definition of an absorption limit is the fact that the dose and passive heat dissipation not only determine the actual temperature increase in tissue but also the active thermo-regulative mechanism of the human body [5], [6]. This highly complex mechanism enables dissipation of whole body temperature increases, and to a much larger extent, local temperature increases. Nevertheless, this mechanism is highly dependent on various factors such as age, health, and which organs are affected. To take these effects also into account the dose concept was extended by defining the absorbed power per unit mass (W/kg) as the critical quantity for safety considerations rather than the absorbed energy per unit mass (J/kg). The rate of the absorbed energy is called Specific Absorption Rate (SAR) and defines the incremental electromagnetic power (dP) absorbed by an incremental mass (dm) within a volume element (dV) with the specific mass density (ρ).

\[
SAR = \frac{dP}{dm} = \frac{dP}{\rho dV}
\]  
(1.1)

The absorption rate can be determined by either assessing the electric field (RMS value of the Hermitian magnitude) in tissue or the temperature increase in tissue (dT/dt).

\[
SAR = c \frac{dT}{dt} = \frac{\sigma}{\rho} |E|^2
\]  
(1.2)

Where c is the thermal constant of the tissue, ρ the mass density and σ the conductivity of the specific tissue. The SAR must be averaged over a certain time period to maintain a relation between absorbed power and the temperature increase in tissue (see Table 1.3).
A correlation between (hazardous) biological effects and the rate of absorbed electromagnetic power was mainly based on exposure situations which lead to whole body temperature increase in animal experiments [7]. This was set as the basis for the maximum tolerable SAR.

Though local temperature increases are seen as less critical this had to be limited as well since nearfield exposure can lead to strong local temperature increases before the whole-body averaged SAR limit is violated. As to local exposure, a minimum averaging tissue mass had to be determined that would reflect the non-hazardous nature of strong superficial heating of the skin. Various averaging volumes were chosen (see Table 1.3, and ref. [8], [9], [10]). The choice of these masses and selecting the cube to describe these masses lacks physical and biological backing and has been subject to ongoing discussions. The cube was selected simply because of the shortcomings of earlier experimental and numerical methods. The fact that a variety of tissues of different specific weights can exist within the averaging volume complicates the evaluation. Better approaches are currently being discussed in the standards organizations.

The assessment of the induced SAR is a difficult task and requires extensive experimental and numerical resources. Furthermore it also requires a modeling of the human body. Therefore, exposure limits were derived which allowed assessment of the absorption rate by solely determining exposure quantities such as incident field strengths. These exposure quantities would be easy to measure using known methods [11].

However, the induced SAR in the human body depends on a variety of complex parameters such as frequency, field polarization and the properties of the exposed biological body. A direct relationship between the exposure and the dose, i.e. the SAR, is difficult to determine, also with regard to the fact that the presence of the body alters the field distribution and in nearfield situations the body directly interacts with the source itself. Exposure limits must therefore include all exposure conditions what approximates the worst possible situation. The derived current exposure limits for the electric and the magnetic field are based on this approach. Maximum absorption occurs in a human being standing upright under plane wave exposure with an incident E-field polarization parallel to the body axes [12]. Under most other exposure situations smaller induced SAR values are found and the derived exposure limits would overestimate the actual absorption. This is especially true when the body or parts of the body are in very close proximity to a radiation source such as a low power handheld mobile telephone. Though the exposure limits would be exceeded, the actual induced SAR would only be a fraction of the SAR limits.

The conclusion was that exposure limits were of no use in assessing the safety of handheld mobile communication devices. The standards bodies were well aware of this problem and introduced exclusion clauses for handheld devices [4]. Transceivers which operated below 1 GHz, and which had an input power of less than 7 watts were excluded from compliance with the basic safety limits. This exclusion clause was adopted worldwide. However the scientific data available in the early eighties did not support the soundness of the 7-watt exclusion clause. These experimental studies had been performed on various transmitters operating at frequencies between 30 and 900 MHz [13], [14], [15]. All through the eighties contradictory conclusions were drawn from numerous studies investigating the absorption induced by sources in the closest nearfield [16], [17], [18]. Only in the early 90s did it become obvious that
<table>
<thead>
<tr>
<th></th>
<th>ANSI95.1-1992</th>
<th>prENV50166-2</th>
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<td>Group 1:</td>
<td></td>
<td></td>
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</tr>
<tr>
<td>whole-body av. SAR</td>
<td>0.4 W/kg</td>
<td>0.4 W/kg</td>
<td>0.4 W/kg</td>
</tr>
<tr>
<td>spatial peak SAR</td>
<td>8 W/kg</td>
<td>10 W/kg</td>
<td>8 W/kg</td>
</tr>
<tr>
<td>averaging time</td>
<td>6 min.</td>
<td>6 min.</td>
<td>6 min.</td>
</tr>
<tr>
<td>averaging mass</td>
<td>1 g</td>
<td>10 g</td>
<td>1 g</td>
</tr>
<tr>
<td>shape of volume</td>
<td>cube</td>
<td>cube</td>
<td>cube</td>
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<tr>
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</tr>
<tr>
<td>Group 2:</td>
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<td></td>
<td></td>
</tr>
<tr>
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<td>0.08 W/kg</td>
<td>0.4 W/kg</td>
</tr>
<tr>
<td>spatial peak SAR</td>
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<td>2 W/kg</td>
<td>8 W/kg</td>
</tr>
<tr>
<td>averaging time</td>
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<td>6 min.</td>
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</tr>
<tr>
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<td>1 g</td>
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<td>1 g</td>
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<tr>
<td>shape of volume</td>
<td>cube</td>
<td>cube</td>
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</tr>
</tbody>
</table>

Table 1.3: Examples of SAR limits proposed in the USA (ANSI95.1-1992 [8]), Europe (preENV50166-2 [9]) and Japan (TTC/MPT [10]) for the frequency range of mobile communications (40 MHz - 6 GHz).

devices with 7 watt input power could substantially violate the basic limits and induce SAR values up to 100 mW/g. The study [19] behind these findings investigated the nearfield absorption mechanism in homogeneous biological bodies caused by a dipole structure at frequencies higher than 300 MHz.

1.1.3 Nearfield Absorption

Most biological tissues of the human head have a high water content with a relative permittivity $\varepsilon_r$ of around 40 and higher. In the frequency range of modern mobile communications (>400 MHz), capacitive coupling is found to be poor whereas inductive coupling becomes dominant.

In [19] a thorough investigation and a detailed description of a nearfield absorption model was performed. It was shown that an absorption mechanism based on inductive coupling is well suited to describe the exposure of a human head to electromagnetic fields in this frequency band. This model is all the more valid for sources generating a dominantly reactive H-field in their vicinity such as helix, loop and wire antennas. In addition, the study concludes that size and shape of the homogeneous body barely show any effect on the absorption mechanism. A simple approximation formula could be derived which allowed estimation of the SAR induced at the body surface as a function of only the incident magnetic field and the electrical parameters of the body [19].

The inescapable conclusion for mobile communication devices was that the specified antenna input power alone cannot provide sufficient information about the induced SAR. The distance between radiating structure and body together with the square of the magnetic field caused by the radiating structure determine the induced SAR in the head. Modern MTEs can be differently designed (antenna types, casing) and held in various positions with respect to the head. Though their input power is approximately the same, the induced SAR in the head could, therefore, vary considerably from device to device.
In the early 90s it became obvious that neither exposure limits nor exclusion clauses could satisfy both industry interests and consumer safety.

1.1.4 Preliminary Tests of Modern MTEs

At around the same time - the end of 1992 - first lawsuits were filed against several manufacturers and service providers which associated the occurrence of brain tumor with excessive use of a mobile telephone. Media response was strong after one of the plaintiffs appeared on CNN's "Larry King Live". The industry was caught by surprise and was not able to prove that its devices were safe as regards the current safety limits. Though the question of cancer is not covered by the safety limits, it would have at least helped to assure public confidence had industry been able to give clear answers regarding compliance with the safety limits. This period of confusion was even briefly reflected in the losses experienced by the major telecommunications companies on Wall Street. Top management's earnest statements that "mobile phones are safe" weren't enough to comfort a concerned public. A couple of months later first preliminary results were published which showed that modern MTE, when tested according to the above-mentioned absorption mechanism, either came close to the safety limits or exceeded them [20]. The assessment system used for those tests had recently been developed with an eye to future compliance tests and was the prototype version of the dosimetric assessment system to be described in more detail later on. Several NMT devices operating at 900 MHz with a specified input power of 1 watt were tested under two different operating positions. The first was a so-called normal or intended use position of the MTE with respect to the head: i.e., the earpiece was at the ear, the microphone as close as possible to the mouth or the cheek. The second test position was a worst case position where, if possible, the antenna touched the head in an area above the ear. The induced averaged SAR was measured by a miniature E-field probe inside a simple human-shaped shell phantom which was filled with a tissue-simulating liquid.

In the period from 1993 to 1995 the prototype version of the dosimetric scanner was further developed and improved. The telecommunications industry was eagerly awaiting a system to test their devices for determining energy deposition in the user's head. This system is now in use in laboratories of all major MTE-manufacturers as well as several service providers in Europe and the United States (DASY2 [21]).

Nevertheless, though an agreement had been reached about the fact that MTE-type approval based on compliance with the basic safety limits is the most sound approach to safeguarding both industry interests and consumers safety, a standardized test procedure for compliance has not yet been defined.

1.2 Compliance Tests

1.2.1 Basic Requirements for Standardization

The basic requirements are that compliance tests cover the exposure of all users under all operational conditions while the numbers of tests should be limited to a minimum. The technical requirements include a high degree of accuracy in assessing the induced SAR in the human head, high degree of reproducibility, the possibility of testing randomly selected devices, time efficiency, and simple test procedures to validate the results. In addition, several other requirements would be advantageous:
ease of use so that tests can be performed by non-specialists, availability to the manufacturer during the design process and credibility in the eyes of both scientists and consumers.

Two different approaches are currently being discussed concerning compliance tests. These approaches are either based on measurements or on numerical computations. The former shows distinct comparative advantages which can be described as follows:

- The actual MTE can be tested during design, or as it comes off the production line. This has become all the more important since preliminary tests have shown that the design plays an important role in the induced SAR due to the above-mentioned absorption model. Detailed numerical modeling of an MTE is critical, requires extensive validation for every model, and is the crucial bottleneck of numerical compliance tests.

- Robot-based measurement systems enable fully automated compliance tests which are time efficient and superior to today's numerical methods. Preliminary tests of MTE using the dosimetric scanner could be performed within 15 minutes.

- In the case of measurements, the entire system, dosimetric scanner and the human model, must be validated only once. Using precision reference sources reproducibility of the SAR assessment can be easily ensured in different laboratories.

- And finally, an experimental compliance test based on an automatic scanner only needs technical personnel to do the tests.

- Measurement approaches have the disadvantage, however, that the human body must be simplified and simulated using shell phantoms filled with tissue-simulating liquids. Numerical methods, on the other hand, have the big advantage that the human body can be simulated in great detail based on discretization of MRI-images of the human body. Numerical methods are superior in assessing the induced SAR in specific tissues. The feasibility and the advantages of a simplification of the body for purposes of compliance testing lies within the scope of this thesis.

Another important requirement is the question of which operating position of the MTE must be tested with respect to the head. In view of the massive growth in users, some health agencies began calling for dosimetric type approval and requiring compliance with safety limits under all operational conditions [22], [23]. Such a "worst case" approach is directly analogous to the commonly applied safety considerations for chemical and physical agents. Some manufacturers and service providers prefer to test under intended use or normal positions only, a position which is described in their user manuals. This is because of their concern that more strict requirements cannot be satisfied with current technology, especially in view of consumer desires to purchase ever-smaller and lighter devices. It is clear that desired growth can only be realized with an attractive high-tech consumer product. The current discussion will most likely lead to the testing of several operating positions.

Finally, the question of a standardized model of the human body or the human head must be addressed. Mobile phones operate in direct proximity to the human
Figure 1.1: Factors determining the SAR in the human head and their inter-dependencies. These interaction factors must be considered and covered by compliance tests.

head. Due to their operational conditions and their small sizes the interaction between the user's head and the radiating structure becomes an important factor for the induced SAR. Next to determining the decisive absorption factors, these interactions must be considered as well when designing a standard model which should cover the exposure of all users under all operational conditions.

1.2.2 Difficulties in Determining the SAR

The main factors that determine the induced SAR in the head of the user are the electrical parameters of the regions of the head mainly affected, the magnetic field strength (i.e. antenna current) and the distance of the radiating structure from the head [19]. The induced SAR caused by mobile telephones depends on various additional factors which must be considered when designing a compliance test.

- **Head and Body:**

  The actual SAR depends on the size and shape of the head. The spatial SAR also depends on complex anatomy of the human head. This includes the role of inhomogeneities in the brain but also such as skin, fat, bone layers of varying thickness. The influence on the averaged SAR must be investigated. This question is extended to the complex anatomy in the ear region, with massive bony structures and air cavities. Another problem is the assessment of actual living tissue parameters and their simulation.

  Next to the anatomical factors external metallic objects such as glasses and jewelry, which are excited as passively radiating structures in direct proximity...
to the head, must be considered. Internal metallic structures such as head or dental implants might also influence the maximal induced SAR-values.

• **Relative Position of the MTE With Respect to the Head:**

A vast number of operating positions exist for the use and positioning of a mobile phone. Furthermore, the antenna mounting on an MTE is usually unsymmetrical regarding the front plane middle axis. Depending on the preferences of the user the device can be operated in the left or the right hand which also leads to big variations in the induced SAR.

• **MTE:**

It must be expected that variations exist among devices of the same model due to manufacturing tolerances and due to variations in the specified input power according to the allowed tolerances. In addition, the design of the antenna, its construction and its mounting site on the device also strongly determine the absorbed SAR. Finally, most MTEs can be operated with the antenna retracted which usually leads to severe drop in radiating efficiency and poses no problem. However, in this case some models use a secondary antenna, usually a helix. Though the radiation efficiency is still smaller than with the primary antenna extended, the reduced distance between the radiating structure and the body and the different H-field distribution could still induce high SAR-values.

In addition, the following peculiarities - which reflect the interaction of the above-mentioned factors, MTE and head - complicate the definition of a standardized test method:

• The highly complex anatomy in the ear region of the head shows various tissues of strongly differing electrical parameters. Small shifts of the device parallel to the ear will lead to strong variations in the assessed maximum SAR and the SAR-distribution. Furthermore, the individually differing outer anatomical shape of the head in this region also causes varying results from user to user due to the small distances.

• The induced SAR is determined by a variety of factors and their interdependencies as displayed in Figure 1.1. Given the operating position of the device with respect to the head and the current and current distribution (on the antenna and the device), the absorbed power can be calculated by H-field coupling. However, in reality the input power and the antenna current are dependent on the operating position. Thus a simple calculation based on all these factors alone would be very difficult if not impossible. Current measurements on devices without the body in close proximity give rise to big uncertainties.

• Uncertainties in determining the averaged maximum SAR should be added as additional safety factors to the measured value. Previous studies showed that the actual averaged maximum SAR values caused by modern MTEs come close to the safety limits [3]. To satisfy both consumer safety and industry interests the uncertainties must be included but should be reduced as much as possible. This entails an exact quantification of all possible factors.

The rationale is often expressed that, since large safety factors have been incorporated in the standards, exceeding these limits under certain circumstances is
acceptable. However, this argument is contradictory to the basic idea of safety standards, since these factors do consider the uncertainty of the extrapolation from animal experiments to the human organism. In addition, they should also incorporate possible “sensitivity variations” among humans. It would therefore not be consistent to apply them to the uncertainty of the exposure as well.

1.2.3 CENELEC/WGMTE Proposal

In 1995 CENELEC, under the mandate of DGIII [24], commissioned a working group for mobile telecommunications equipment (WGMTE) to develop a standard for compliance tests of MTEs. The proposal, a draft of which was published in early 1996 [25], can be summarized as follows:

- Compliance with the safety limit of 2 mW/g averaged over 10 g of tissue and 6 minutes of exposure according to prENV 50166-2 [9].
- Compliance must be proven for all users.
- Several operating positions must be tested. All positions must comply with the basic safety limits.
- All assessment uncertainties must be added to the measured SAR.

The devices being tested must operate with available maximum power and with antennas extended as well as retracted. Four operating positions which ought to be tested are defined. One position can be described as the intended operating position (see Figure 1.2) and three as additional operating positions (Figure 1.3). This proposal attempts to reach a compromise between a worst case approach and the normal or intended use of a mobile telephone usually described in the user’s manuals. The four positions with respect to the head (Figures 1.2 and 1.3) are defined by angles between two distinctive planes in the head and on the device. It is a straight-forward method, though the choice of the test positions is rather arbitrary [25].

This WGMTE/CENELEC proposal is the first which includes test positions in the standard. In a further approach, a description of a standard model to be used will be added. The latter will rely on the scientific bases of this study.

1.2.4 Conclusion

An experimental method based on an automated dosimetric scanner seems best suited to compliance testing. However, the measurement approach has the one disadvantage that the human body must be simulated and considerably simplified. A miniaturized E-field probe can only be freely positioned in a liquid. Based on the results from [19], an early phantom model was manufactured which consisted of a thin fiber-glass shell resembling in shape a human torso with head and which was filled with a (brain) tissue-simulating liquid. This phantom was regularly criticized for being an over-simplification of the anatomically complex human head and therefore, its accuracy in assessing the maximum averaged SAR was questioned.

For a standard model for compliance tests it is not necessary or perhaps not even reasonable to use a detailed head model of a specific human. It should rather be simplified such that all variations mentioned above might automatically be taken into account. The requirements can be summarized as in the following:
The averaged SAR induced in the model should never be smaller than the maximum averaged actual value in the human head regarding all possible interaction factors as mentioned above.

The over-estimation of the maximum averaged SAR value should be as small as possible.

The uncertainty of the assessment of the maximum averaged SAR value should be as small as possible.

The behavior of the device in proximity to the model regarding feedpoint impedance and current distribution should resemble the actual situation as closely as possible. This must be valid under various operating positions of the device with respect to the head.

Reproducibility must be guaranteed. Tissue parameters and shapes must correspond to reality and no correction factors should be necessary. Otherwise, it would be necessary to validate the reliability of the measurement method for every new device or antenna type.

This thesis tries to set the scientific bases for the design for compliance tests based on the above factors. The problem of an accurate modeling of the absorption
regarding compliance testing of MTE can be split into three parts which correspond to Part II, III and IV in this study:

- **Increase in Measurement Accuracy (Part II)**
  The first addresses the question of major intrinsic uncertainties regarding the dosimetric assessment method which is used. A preliminary uncertainty estimation revealed that known calibration methods in lossy media for miniaturized E-field probes and the determination of the electrical parameters of the simulating liquid introduce large uncertainties (up to ±30%). In view of the significance of the safety limits for mobile communications, additional uncertainties would imply unnecessary restrictions to future MTEs.

- **Human Head Modeling (Part III)**
  The second deals with the human anatomy in the head regarding the maximum induced spatial SAR. The influence of inhomogeneities in the brain as well as those in the entire head are investigated. The role of size and shape are determined. If feasible the complexity of the human head must be reduced for compliance testing. The errors due to these simplifications must be quantified.

- **Considering Additional Factors with a Model of the Head (Part IV)**
  The last part addresses additional effects influencing the induced SAR such as the hand of the user holding the device, external and internal metallic structures and absorption in head regions where lower absorption is expected. An overall error estimation should finally identify the most critical factors in assessing the maximum averaged SAR for compliance tests.
References


Part II

Increase in Measurement Accuracy
Leer - Vide - Empty
Chapter 2

Broadband Calibration of E-Field Probes in Lossy Media

Abstract A broadband calibration procedure for E-field probes has been developed that minimizes the overall uncertainties inherent in E-field measurements in lossy dielectric liquids. The analysis of the calibration requirements shows that probes which are symmetrical with respect to their axis greatly facilitate accurate calibration, since the calibration procedure can be divided into several discrete steps. Such a procedure is presented and analyzed with respect to its uncertainties. Absolute calibration is performed at three frequency bands utilized in Europe for mobile communications (450 MHz, 900 MHz and 1.8 GHz) and in different tissue-simulating liquids. The parameters obtained are verified by numerical simulations of the probe in the surrounding media. Such simulations allow the assessment of some of the calibration parameters with sufficient accuracy in cases where the experimental determination would be too tedious and time-consuming.

2.1 Introduction

In view of the phenomenal growth of the mobile communications market, the telecommunications industry has lately recognized the need to test its mobile telephones for compliance with safety limits. The current safety limits for the frequency range of mobile communications are defined by the maximum tolerable absorbed power per tissue mass in W/kg, known as the specific absorption rate (SAR). The local SAR can be determined experimentally by measuring either the induced electrical field strength $E$ (RMS value of the Hermitian magnitude) or the temperature rise $\partial T / \partial t$ caused in the tissue by the absorption.

\[
SAR = \frac{\sigma}{\rho} E^2 = c \frac{\partial T}{\partial t}
\]  

(2.1)

where $\sigma$ is the conductivity, $\rho$ is the mass density and $c$ is the specific heat of the tissue at the site of measurement.
Since measurements using thermal probes do not provide an adequate degree of efficiency and sensitivity for compliance-testing of consumer products, research up until today has been focused on E-field probes inside tissue-simulating media.

The original design of a miniaturized isotropic E-field probe for use in tissue-simulating liquids goes back to Bassen et al. [1]. In [2] and [3] a new probe design with significantly improved performance characteristics is presented.

In view of the significance of and difficulties involved in accurate calibration, surprisingly little has been published so far about broadband calibration of isotropic E-field probes in dielectric materials. In [4] a calibration procedure in an S-band waveguide at a single frequency of 2.45 GHz is described. However, the calibration uncertainties due to the dependence of the probe sensitivity on polarization, frequency, dielectric parameters of the surrounding media and spatial resolution have only been marginally addressed. If these effects are not carefully considered, the measurements errors can easily be in the range of 3 to 6 dB or larger. Especially in view of the limited leeway possible with respect to the safety limits of modern telecommunications equipment [5], such uncertainties are not tolerable for dosimetric type approval setups.

In this chapter the error sources which depend on the design of the E-field probes are discussed. Based on these considerations a broadband calibration procedure for the dosimetric probe previously presented in [2] was developed with the objective of minimizing the uncertainties for dosimetric assessment.

### 2.2 Calibration Requirements

E-field probes with isotropic response are achieved by the orthogonal positioning of three sensors, each sensitive to one E-field component. Short dipoles or small E-field-sensitive crystals have these characteristics. In miniaturized E-field probes small dipoles equipped with diode rectifiers are generally used, since they offer the greatest sensitivity and have a linear response over a wide frequency range. Phase information is not required, since SAR is proportional to the Hermitian magnitude. For use in liquid media the sensors must be protected, i.e., encapsulated. This usually affects the performance of the probe considerably.

The probe (SPEAG Model ET3DV4) for which the calibration is demonstrated is shown in Figure 2.1. It consists of three small dipoles (3 mm in length) directly loaded with a Schottky diode and connected via resistive lines to the data acquisition electronics. The core which holds the ceramic substrates as well as the outer shell are made of the synthetic microwave material STYCAST0005 with a permittivity of 2.54 and a loss tangent of 0.0005. In its center the optical multifiber line is inserted, enabling contactless surface detection.

The basic requirement for calibration is that the output signals of three orthogonally positioned sensors be evaluated in such a way that the reading corresponds to the SAR at the measurement site in the absence of the probe. Unfortunately, the relation between the field and the sensor signals depends on several factors:

- design and construction materials of the probe;
- electrical properties of the surrounding media;
- direction and polarization of the field;
Figure 2.1: Tip of the E-field probe. The tip encapsulation has been removed. One 3 mm long dipole and the diode can be seen. In the center of the core is the opening for a built-in optical proximity sensor.

- field gradient at the measurement site;
- RF characteristics of the antenna, the rectifying element and the transmission line;
- higher order modes or different reception modes in the probe;
- sensitivity of the rectifier;
- characteristics of the evaluation circuit for the rectified signals.

The calibration essentially attempts to describe these effects quantitatively so that correct SAR values can be obtained under various measurement conditions. Furthermore, it is important to know the absolute uncertainty and the validity range of the calibration. However, the first four factors have scarcely been addressed in previously applied calibration techniques. These factors are briefly discussed in the following:

*Influence of the probe material on the field:* Any dielectric material around electric dipoles generally alters the local signal strength inside the probe. For probe constructions as shown in Figure 2.1 the influence on E-field components normal to the probe axis will be different from that on E-field components parallel to the probe axis. Furthermore, this difference in sensitivity depends on the surrounding medium. This results in poor isotropy in planes which are aligned to the probe axis. For example, a deviation from isotropy of less than 0.2 dB is easily achievable around the probe axis (i.e., E-field polarization normal to the probe axis), owing to symmetry. However, for E-field polarizations in planes aligned to the probe axis, the deviation
is larger than 1.5 dB in air and reduces to 0.6 dB in brain-equivalent tissue [2]. In [3] several methods are presented to compensate for these effects. In any case, the directivity characteristics of the probe will be different in different media. In addition, these effects are frequency-dependent in lossy media.

**Effects in strongly non-homogeneous fields:** Special considerations are needed if the field significantly changes within the probe's dimensions, e.g., at higher frequencies inside lossy material. Therefore, a calibration reference point in the probe is defined. However, the sensor response may depend significantly upon the probe's alignment with respect to the direction of the field gradient. In addition, each field component is measured at slightly different locations, due to the spacing between the sensors. These effects result in an increased deviation of isotropy and must be carefully analyzed.

**Influence of material discontinuities:** In the immediate vicinity of material discontinuities the dielectric body of the probe has a significantly different effect on the field than within homogeneous material. For example, an increase in sensitivity can be observed when the probe approaches the surface of a shell phantom (see Figures 2.5 to 2.8), since the probe's tip is only barely surrounded by liquid. In this case the calibration performed for homogeneous materials is not valid.

To keep the number of calibration parameters and calibration measurements low, it is crucial to separate the above-listed influences and quantify them individually. Since any calibration is principally valid only for the special conditions under which the calibration has been performed, an analysis of uncertainty must be performed for the conditions under which the probe is actually used. For axis-symmetrical probes it is possible to calibrate the probe in a three-step approach, as shown in the next section. Furthermore, the proposed setups for calibration also closely correspond to those of the dosimetric assessments performed with the scanner, as described in [2].

### 2.3 Three-Step Calibration Approach

A three-step approach is possible if the calibration factor can be separated into three independent factors $f_i(V_i)$, $\eta_i$ and $\gamma_i$:

$$|E|^2 = \sum_{i=1}^{3} |E_i|^2 = \sum_{i=1}^{3} \frac{f_i(V_i)}{\eta_i \gamma_i}$$  \hspace{1cm} (2.2)

where $V_i$ is the rectified signal from the sensor elements.

**a) $f_i(V_i)$:** In most cases $V_i$ needs to be linearized, because of the nonlinear response of the output dependent on the rectifier (diode compression) and evaluation circuit. This is a function of the magnitude of the rectified signal only and independent of the RF-transmission to the diode. $V_i$ is monitored during a power scan of the exciting field. The linearization function is evaluated, so that $f_i(V_i)$ is proportional to the square of the exciting field strength. When the rectifying elements and evaluation channels are identical, the same function $f_i(V_i) = f_i(V_i)$ can be used for all sensors. In the case of amplitude-modulated signals, the timing characteristics of the evaluation circuit must also be taken into account. For pulsed signals with a known crest factor, a simple correction formula can be given; for arbitrary modulations, however, a more sophisticated signal analysis is necessary.

**b) $\eta_i$:** These factors describe the relation between the linearized signal of a single sensor $f_i(V_i)$ and the field component in the direction of the sensor $(\mu V/(V/m)^2)$ in
Figure 2.2: Setup for calibration in wave guides, i.e., in air. Measurements a and b to calibrate the power meters $P_1$ and $P_2$ with respect to the high-precision meter $P_3$. Probe measurements c and d with different distances to load. By adjusting the amplification to keep $P_1$ constant, the wave guide input power is equal to the power reading $P_3$ plus the attenuation of the calibrated attenuator.

Aside from the dipole length, they depend on the probe's materials, the sensor's positioning and the RF characteristics of the sensor components. If detector diodes are used as rectifying elements, the parasitic capacitance, which is generally not precisely specified, influences the RF behavior. Therefore, the factors $\eta_i$ will be different for each sensor, even if the sensors are positioned in a symmetrical fashion. These factors can be assessed by standard probe calibration procedures (see Section 2.4). All error sources (isotropy, frequency linearity) must be investigated during this calibration in order to assess the calibration factors for an average measurement situation and to evaluate the error and validity range of the calibration. For broadband E-field probes, the calibration factors are independent of the frequency over a wide range (2 to 3 magnitudes) and can thus be considered to be constants.

c) $\gamma_i$: These factors describe the ratio of the probe sensors' sensitivity in different media to their sensitivity in air, i.e., $\gamma_i = 1$ for air. These usually depend both on the surrounding material and the frequency, and on the materials and the design of the probe. In other words the (time-consuming) assessment of these calibration factors (see Sections 2.5 and 2.6) need only be done once for each probe type, and not for each individual probe. It is further necessary to reassess the deviation from isotropy in liquid, since it may differ from air (see Section 2.2). In the case of symmetrical sensor positioning they will in fact be identical for each sensor ($\gamma_i = \gamma$). $\gamma$ will hereafter be called the “conversion factor”.

The separation of the calibration factor in a probe and sensor-dependent factor $\eta_i$ and a probe type and situation-dependent conversion factor $\gamma_i$ is an approximation that is based on various assumptions:

- The variation of the dipole impedance caused by the surrounding medium is
the same for all sensors.

- The local E-field distribution in the area of the sensors inside the probe only depends in magnitude on the surrounding medium. In symmetrical probes this fact can be regarded more leniently.

- The differences (manufacturing tolerances) among probes of the same type are small.

The validity of these assumptions depends largely on the probe's design. Measurements and simulations of our probes have established the feasibility of this calibration procedure. In the following the setups and procedure are described in detail with which the factors \( \eta \) and \( \gamma \) have been determined.

### 2.4 Calibration in Air

To calibrate the probe in air, a well-defined measurement volume with a known and largely homogeneous electrical field is necessary. Since reference probes are usually not accurate enough, the field strength must be determined from power measurements. Depending on the frequency, different setups were used to calibrate the probe in air:

a) For frequencies over 1 GHz standard wave guides R22 and R26 were used with the setup according to Figure 2.2. The probe was rotated around its axis with a positioning accuracy at the probe tip of better than ±0.1 mm. By using high-precision components (loads, lines, adaptors) and error compensation methods, an absolute accuracy of better than ±5% was achieved. The linearity over different frequencies and wave guides is better than ±2%. The probe produced reflections of 1.6% in the R22 wave guide and of 2% in the R26 wave guide.

b) At frequencies below 1 GHz a TEM cell (ifil10) with rectangular cross section was used. The field gradient at the calibration location in the center over the septum was less than ±2%/cm. By taking into account all error sources in the field calculation an absolute accuracy of not better than 10% can be achieved. However, the agreement with the wave guide measurement was within ±3% and the linearity from 30 MHz to 900 MHz was within ±2%. Probe reflections were found to be negligible.

c) In the waveguide and TEM-cell measurements, the isotropy around the probe axis can only be assessed. To measure the isotropy in all directions, the following nearfield setups were applied. At lower frequencies the field in the symmetry plane of symmetric standard dipoles was used (similar to the setup of Figure 2.3). At higher frequencies, the center point over an open wave guide was chosen as the measurement location. These measurement setups permit the assessment of the isotropy in all directions. The same setups can also be used to determine the deviation from isotropy inside different media [2].

### 2.5 Experimental Determination of Conversion Factor \( \gamma \)

#### 2.5.1 Setup

To determine the conversion factor \( \gamma \), a well-defined SAR distribution is needed inside the dielectric material for which the probe must be calibrated. The local
SAR values can be experimentally measured using small thermal probes according to equation (1) at high power levels. Furthermore, setups are preferred which allow the computation of the field inside the dielectric material either analytically or by numerical simulations. One way would be to use a dielectric slab in a rectangular waveguide. Although the induced fields are well defined and can be easily determined if the emergence of spurious higher modes can be sufficiently suppressed, this setup is very narrow-banded.

Another setup used in [6] is the simulation of a dielectric half-space, which is exposed to a $\lambda/2$ dipole aligned parallel to the surface of this half-space (Figure 2.3). For this configuration accurate results can be achieved by computer simulations since the SAR on the axis is largely proportional to the square of the antenna feedpoint current and not to the output power or to the incident E-field, i.e., does not require integrals in closest proximity to the gap. The drawback is that the feedpoint current can only be experimentally assessed with fairly large uncertainties of no better than $\pm 10\%$.

This setup was nevertheless chosen since:

- It is easy to set up and handle.
- It provides much greater flexibility, since a broad frequency range can be covered by the same setup.
- It is a good representation of the test situation implemented for dosimetric assessments of mobile communications devices [2].

Figure 2.3: Experimental setup. On the top is the plexiglass box filled with the simulating liquid. At the bottom is the dipole. The temperature probes and the E-field probe are positioned directly above the dipole feedpoint.
<table>
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<th>f [MHz]</th>
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<th>$\gamma$</th>
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<tr>
<td>450</td>
<td>47.0 ± 5%</td>
<td>0.43 ± 6%</td>
<td>6.7 ± 10%</td>
</tr>
<tr>
<td>900</td>
<td>40.0 ± 5%</td>
<td>0.87 ± 6%</td>
<td>6.0 ± 10%</td>
</tr>
<tr>
<td>1800a</td>
<td>40.5 ± 5%</td>
<td>1.65 ± 6%</td>
<td>4.8 ± 10%</td>
</tr>
<tr>
<td>1800b</td>
<td>41.0 ± 5%</td>
<td>1.25 ± 6%</td>
<td>4.8 ± 10%</td>
</tr>
</tbody>
</table>

Table 2.1: Dielectric properties and conversion factor $\gamma$ of brain tissue-simulating liquids at the tested frequencies of 450 MHz, 900 MHz and 1.8 GHz. At 1.8 GHz, two liquids were used: sugar-water solution (a) and butyldigol-water solution (b).

The half-space was experimentally simulated by an acrylic glass box filled with the lossy dielectric liquid. The size of the acrylic glass box was 800x600x200 mm$^3$, the thickness was 4 mm. The standard dipoles were placed parallel to the dielectric surface at distances that were small compared to the dimensions of the box and to the distance from the floor. The floor was lined with absorbers.

2.5.2 Brain-Simulating Liquids

The dielectric data for brain tissue available in the literature varies considerably, i.e., up to ±25%, whereby uncertainties and variations in the values for human brain tissue are not given. The basis for our studies was the most recent data for living tissues. In [7] the mean values for grey and white matter at 900 MHz are $\varepsilon_r = 43$ and $\sigma = 0.85$ mho/m. In [8] white matter at 450 MHz is determined to have $\varepsilon_r = 48$ and $\sigma = 0.6$ mho/m. Liquids which allow the simulation of similar brain parameters at 450 MHz and 900 MHz and which are inexpensive and easy to handle consist of sugar, water, NaCl and hydroxyethylcellulose (HEC) [9], [8]. For brain tissue at 1.8 GHz, two different liquids were used. The first (1800a) was based on a simple sugar-water solution without any salt (free ions). Nevertheless, the conductivity of such a solution was still higher than the brain tissue parameters in [7] (Table 2.1). For grey matter the corresponding values would be $\varepsilon_r = 41$ and $\sigma = 1.45$ mho/m. The reason is that the bound sugar-water complexes begin to determine the conductivity of the liquid at frequencies higher than 1 GHz. Therefore, sugar was replaced by butyldigol (2-(2-butoxyethoxy)ethanolbutyl) which, when dissolved in water, shows lower conductivity values. At the same $\varepsilon_r$ value, conductivity could be reduced from 1.65 to 1.25 mho/m (1800a). The mean values for grey and white matter according to [7] are $\varepsilon_r = 41$ and $\sigma = 1.15$ mho/m.

The electrical parameters were measured by an open coaxial method using the HP 85070A Dielectric Probe Kit. To verify the open coaxial method, we determined the electrical parameters using the slotted-line method. The agreement was within 4%. Table 2.1 gives an overview. The temperature dependence of the liquid's parameter was also checked. In the temperature range between 15° C and 30° C, a change of 5% in conductivity was measured.

The specific thermal constant $c$ was determined using a simple calorimetric procedure with an accuracy of better than ±4%. For brain tissue-simulating liquids, $c$ was determined to be 2.85 J/K/g ±4% with a specific density ($\rho$) of 1.30 g/cm$^3$ ±1%. Comparison of these values for $c$ with data from the literature shows considerable agreement. Gucker et al. measured 2.90 J/K/g for a similar aqueous sucrose
Figure 2.4: Temperature increase measured during RF exposure to a 900 MHz field. The power input was 43 dBm. d = dipole distance from the simulating liquid; a = sensor distance from the acrylic glass bottom.

The value for the brain tissue-simulating liquid which uses the sugar substitute buthyldigol is 3.58 J/K/g (ρ = 0.98 g/cm³ ±1%).

2.5.3 Temperature Probe

The measurement of the local SAR by temperature probes has the advantage that small sensors with an active area below 1 mm² are readily available. Optical probes or thermistor probes with high resistive lines [11] provide the necessary field immunity. In the calibration process the limited temperature sensitivity of these sensors can be overcome by applying high power.

In this study, a new non-metallic temperature measurement system was used to measure the temperature increase. The probe is based on an NTC temperature sensor (negative temperature coefficient) connected to four resistive lines [12]. The noise level of this system is about 100 times less than that of the two comparable optical devices on the market (±0.001°C averaged over 0.1 s; < ±0.1 mK/s for 10 s exposure). The temperature increase in the liquid due to exposure to a 900 MHz field at two distances from the body is shown in Figure 2.4. After every exposure period, the liquid was stirred until thermodynamic equilibrium was reached.

Thermodynamic dissipation processes are critical because of the large tempera-
Figure 2.5: SAR assessed by E-field (circles) and temperature measurement (squares) at 450 MHz in brain-simulating liquid ($\epsilon_r = 47.0$ and $\sigma = 0.43$ mho/m). The SAR was normalized to 1 watt input power. Due to the inefficient coupling at 450 MHz, the closest possible distance of 12 mm between dipole and body was chosen. The conversion factor $\gamma$ was determined to be $6.7 \pm 10\%$. The "wavy" shape of the absorption curves can be explained by reflections at the liquid-air boundary. This phenomena disappears at higher frequencies.

Temperature gradients induced and were assessed by evaluating different time intervals. They were found to be negligible within the first 10 seconds for sugar-water solutions. The solution based on butyldigol had a considerably lower viscosity, so that the evaluation had to be reduced to an interval of 5 seconds.

A robot positioned the temperature probe in the liquid with an accuracy of better than $\pm 0.2$ mm. The disturbance caused by the probe holder has been shown to be negligible.

2.5.4 Results

The conversion factor $\gamma$ was determined by comparing the values of the temperature probe and of the E-field probe on points along the line which is normal to the Plexiglass bottom and above the dipole feedpoint (Figure 2.3). These measurements were repeated at different power levels, at different distances of the RF source from the body and at various frequencies (Figures 2.5 to 2.8). The conversion factor was assessed by a least-square procedure considering all measured values. The results for
Figure 2.6: SAR assessed by E-field (circles) and temperature measurement (squares) at 900 MHz in brain-simulating liquid ($\varepsilon_r = 40.0$ and $\sigma=0.87 \text{ mho/m}$). The SAR was normalized to 1 watt input power. The dipole distances from the body were 20 mm, 30 mm and 50 mm. The conversion factor $\gamma$ was determined to be $6.0 \pm 10\%$.

The accuracy of the conversion factor $\gamma$ is mainly determined by the uncertainties in determining the correct electromagnetic and thermal properties of the tissue-simulating liquid. Using the “slot line” and “open coaxial” methods to determine the conductivity of the liquid results in an uncertainty of about $\pm 6\%$. The specific heat of the liquid can be assessed to an accuracy of about $\pm 4\%$. The E-field probe yields another $\pm 5\%$ and the temperature probe $\pm 3\%$. Since these uncertainties and those of the positioning of the E-field probe ($\pm 1\%$) and of the temperature probe ($\pm 2\%$) as well as that of the power meters ($\pm 1\%$) can be considered to be statistically independent, the total uncertainty calculated based on root-mean-square is less than $\pm 10\%$.

2.6 Numerical Determination of Conversion Factor $\gamma$

To obtain a greater insight into the behavior of the conversion factor, two numerical program packages based on two different techniques were used. This had the advantage of cross-validating the modeling and took advantage of the different strengths...
Figure 2.7: SAR assessed by E-field (circles) and temperature measurement (squares) at 1.8 GHz in brain-simulating liquid ($\varepsilon_r = 40.5$ and $\sigma = 1.75$ mhos/m). The SAR was normalized to 1 watt input power. The dipole distances from the body were 10 mm, 20 mm and 30 mm. The conversion factor $\gamma$ was determined to be $4.8 \pm 10\%$.

of each method.

The first technique applied was the 3D MMP software package. This is a frequency domain boundary technique suited for 2D and 3D scattering problems within piecewise linear, homogeneous and isotropic domains. Details are given in [13] and [14].

The second software package, "MAFIA" is based on the finite integration technique (FIT). This technique is conceptually slightly different from the FDTD but results in the same numerical scheme. The open domains are bounded by second-order Mur absorbing boundary conditions. Details are given in [15] and [16].

2.6.1 Modeling of the Probe

To study the field distribution inside the probe depending on the electrical parameters of the surrounding media, different discretizations of the E-field probe were chosen. A transversal cut of the probe and a perspective view with the location of the dipoles inside the probe is shown in Figure 2.9. Different discretizations with increasing complexity have been compared:

a) The simplest numerical representation of the probe is a simple homogeneous,
Figure 2.8: SAR assessed by E-field (circles) and temperature measurement (squares) at 1.8 GHz in brain-simulating liquid ($\varepsilon_r = 41.0$ and $\sigma=1.25$ mho/m). The SAR was normalized to 1 watt input power. The dipole distances from the body were 10 mm, 20 mm and 30 mm. The conversion factor $\gamma$ was determined to be 4.8 ±10%.

lossless cylinder 6.8 mm in diameter with the electrical properties of STYCAST 0005. The length of the cylinder is 15 mm, which has proven to be long enough to study the fields induced in the probe tip. MMP could only be used for this simple homogeneous model of the probe. The modeling with MMP required about 550 matching points at the boundary of the two domains and 9 multipole expansions. As expected, the maximum errors (<10%) appeared on the matching points at the corners of the probe. In order to minimize these errors and to use a minimal number of expansion functions, the edge of the probe was slightly rounded. About 90,000 voxels were sufficient with MAFIA to model the whole computational domain of which about 5,000 were used for the probe itself. Assuming open boundary conditions and that the whole computational domain is of a lossy material, problems will occur at the outer boundaries. The influence of possible reflected waves was assessed to be less than 3%.

b) A more complex model simulated the optical fiber in the center of the core. The fiber was discretized as a smaller homogeneous cylinder 1 mm in diameter with a relative permittivity of 5 in the center of the STYCAST cylinder.

c) Additional details were incorporated in a further model by the modeling of the three air holes (Figure 2.9).
d) In the most complex model, three ceramic sheets on which the dipoles and lines are printed were simulated as well. This involved a discretization with 210,000 voxels, about 37,000 of which were needed for the probe itself.

The dipoles were not simulated in any of the models. The conversion factor can be calculated by integrating the electric field over the length of the dipoles (ds), first with biological tissue surrounding the structure and then with air surrounding it:

$$
\gamma = \frac{\sum_{i=1}^{3} (\int_{Dipole_i} \vec{E} \cdot ds (in \ tissue))^2}{\sum_{i=1}^{3} (\int_{Dipole_i} \vec{E} \cdot ds (in \ air))^2}
$$

(2.3)

This was performed for an incident plane wave, with the Poynting vector parallel to the probe axis and coming from the front. The E-field at the location of the dipole center in the absence of the probes was chosen to be 1 V/m in both cases.

### 2.6.2 Results of the Simulations

The simulations with the simple homogeneous models were performed for different dielectric properties of the probe's surrounding medium. Figure 2.10 shows the dependence of the probe's conversion factor as a function of the relative permittivity. The frequency of the excitation was set to 900 MHz. These calculations were made for absorbing biological tissue and for non-absorbing tissue.

The conductivity of the lossy material corresponds to the value used for the experimental investigations (0.88 mho/m, see below). The influence of the conductivity of the lossy material becomes less important for a larger real part of the complex permittivity. Within a wide range of relative permittivities (of biological tissue), the conversion factor or, in other words, the sensitivity of the probe changes by less than 10%. This is even true when changing the conductivity of the biological tissue.
within a certain range. However, for small real parts of the complex permittivity, the influence of the conductivity on the conversion factor is large. A comparison of the results of the two methods (MMP and MAFIA) reveals that the difference is less than 1% for the non-absorbing material and between 1% and 3% for the absorbing material.

In Figure 2.11, the results for different MAFIA models are compared. The results of the homogeneous model are the same as discussed before. The conductivity was again chosen to be 0.88 mho/m and the frequency 900 MHz. The effect of the optical fiber inside the probe can be neglected.

For the more complex model with an optical fiber, ceramic sheets and air holes, $\gamma$ is about 6% lower than for the homogeneous model. Additional simulations which neglected the ceramic sheets or the air holes revealed that the air holes are responsible for the drop in the conversion factor.

In Figure 2.12, the frequency dependence of $\gamma$ is shown. The homogeneous models simulated with MMP and MAFIA are in close agreement with each other. Again, the values for the more complex modeling are slightly lower than those for the homogeneous modeling. In contrast to these findings, the experimentally determined conversion factors are larger than those of the simulations, which clearly demonstrates the limitations of this approach.
The reasons lie in the fact that any modeling involves many simplifications of the real probe, e.g., electrical parameters of the probe material were not measured, but were taken from the literature. The most important effect is most likely the change of the dipole capacitance, which depends on the surrounding media and could not be considered in the simulations. This effect, however, is expected to be more significant for probe designs in which the dipoles are positioned closer to the surrounding medium.

2.7 Conclusions

A procedure has been presented which allows an absolute calibration for SAR measurements with an accuracy of better than ±10% for the described condition which closely corresponds to that of the actual dosimetric assessments performed with the scanner described in [2]. Thus, further considerations with regard to polarization are not required. If such studies are needed, the techniques described can be used (see also [2]). Numerical techniques have proven to be adequate to assess the conversion factor $\gamma$ if a precision of about ±20% is sufficient for this probe. For other probe designs, the uncertainties of the numerically determined conversion factors might be considerably larger.
Figure 2.12: Experimentally assessed conversion factor for brain tissue-simulating liquid in comparison with the values obtained from numerical simulation for homogeneous (empty symbols) and non-homogeneous (filled symbols) modelings of the probe.
References


Leer - Vide - Empty
Part III

Head Modeling
Leer - Vide - Empty
Chapter 3

The Dependence of EM Energy Absorption upon Human Head Modeling at 900 MHz

Abstract- In this chapter the dependence of electromagnetic energy absorption at 900 MHz in the human head on its anatomy and its modeling are investigated for RF-sources operating in very close proximity to the head. Different numerical head models based on MRI scans of three different adults are used with voxel sizes down to 1 mm$^3$. Simulations of the absorption are performed by distinguishing the electrical properties of up to 13 tissue types. In addition simulations with modified electric parameters and reduced degrees of complexity are performed. Thus, the models greatly differ from each other in terms of shape, size and internal anatomy. The numerical results are compared with those of measurements in a multi-tissue head model and two homogeneous phantom model of different shapes and sizes. The results demonstrate that size and shape are of minor importance. Although local SAR values depend significantly on local inhomogeneities and electric properties, the volume-averaged spatial peak SAR obtained with the homogeneous models only slightly overestimates that of the worst-case exposure in the inhomogeneous head models.

3.1 Introduction

In the past few years there has been an increase in public concern about possible health risks from the electromagnetic (EM) energy emitted by handheld mobile telecommunications equipment (MTE). The first step necessary for an assessment of the potential risks is to analyze and quantify the EM field induced in the various tissues of the human head by the use of a handheld cellular phone.

Two kinds of dosimetric assessments are of special interest, depending on the emphasis placed on the assessment's endpoint. Detailed information on the distribution of the induced electric and magnetic field strengths inside the various tissues of the head is a prerequisite to designing and performing the most meaningful biological
experiments. On the other hand, an efficient, accurate and reliable assessment is needed to implement a standardized compliance test of MTE for basic safety limits.

To analyze the possible range of variations of the field strengths induced in the various tissues requires an extensive effort, since the local field strengths strongly depend on a large number of parameters, such as:

- operational frequency and antenna input power,
- position of the device with respect to the head,
- design of the device,
- the outer shape of the head,
- the distribution of the different tissues within the head,
- the electric properties of these tissues.

The last three factors are different for various individuals and can even change with time. For example, the outer shape depends on the individual profile and on any movement of the mouth or the eyes. The electric parameters of a human body vary with levels of physical and metabolic activity, health, and age. The variations in all these properties lead to a spread in the analyzed absorption distribution. A strategy on how to obtain scientifically valuable information from this large parameter range has not been worked out yet. However, it is clear that such an approach would not be suited for a type-approval procedure, which should be time-efficient, cost-effective and of utmost reliability. Simplifications are therefore required.

The current basic safety limits applicable for mobile communications equipment are defined in terms of the specific absorption rate (SAR). Three different limits
are defined: 1) a whole-body average SAR, 2) a local peak SAR and 3) a specific absorption (SA), which limits the power of short pulses. 1) and 2) must be averaged over a defined period of time. In the case of an MTE operating at frequencies above 300 MHz, the absorption affects only those parts of the body which are close to the device. Hence, the most critical value is the local peak SAR limit. In Europe the basic limit set for the general public is 2 mW/g averaged over a volume equivalent to 10 g and a period of 6 minutes [1]. The ANSI/IEEE standard [2] defines a stricter limit for an uncontrolled environment of 1.6 mW/g averaged over a volume of 1 g and a period of 30 minutes.

The objective of a type-approval procedure is therefore to assess the maximum spatial peak SAR which the specific device being tested would induce among all users in various operational positions. The system currently most often used for testing MTE is the measurement setup using a homogeneous anatomically-shaped phantom filled with a liquid simulating brain tissue [3], [4]. The rationale behind this approach comes from the energy absorption mechanism in the close nearfield of antennas [5], which concludes that the most determining parameters for volume-averaged values are the time-averaged antenna input power, operational frequency, design of the device and its position with respect to the head, and to a much lesser extent, on the physical properties of the head.

In this chapter the effects of head properties, such as size, shape and inhomogeneities on the absorption are studied to evaluate the applicability of homogeneous head phantoms for compliance tests.

### 3.2 Numerical and Experimental Techniques

#### 3.2.1 Simulation Technique

In electrodynamics by far the most flexible way to investigate effects which depend on multiple parameters is with computer simulations, since geometry and domain properties can be easily varied. Many numerical techniques exist for the analysis of complex nearfield scattering problems, whereby the Finite-Difference Time-Domain (FDTD) technique [11] has proven to be the most efficient method for studying absorption in strongly inhomogeneous bodies. FDTD is currently used by various groups to study the absorption in the human head from mobile phones (e.g., [12], [8], [13], [10]) and to test novel antenna designs (e.g., [14], [15]).

The applied, commercially available code MAFIA [16] is based on the Finite

---

### Table 3.1: Three different head models for computer simulations

<table>
<thead>
<tr>
<th></th>
<th>M1</th>
<th>M2</th>
<th>M3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head volume</td>
<td>4.44 dm³</td>
<td>3.35 dm³</td>
<td>4.26 dm³</td>
</tr>
<tr>
<td>Tissues</td>
<td>13</td>
<td>120</td>
<td>12</td>
</tr>
<tr>
<td>Computational space</td>
<td>175x230x226 mm</td>
<td>159x208x201 mm</td>
<td>159x206x249 mm</td>
</tr>
<tr>
<td>Voxel size</td>
<td>(1 mm)³</td>
<td>(1.075 mm)³</td>
<td>(1.875 mm)²x3 mm</td>
</tr>
</tbody>
</table>
Table 3.2: Tissue parameters for the FIT simulations

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$\varepsilon_r$</th>
<th>$\sigma$ [mho/m]</th>
<th>$\rho$ [kg/m$^3$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>bone (M1, M2, M3, M4)</td>
<td>20.9</td>
<td>0.33</td>
<td>1850</td>
</tr>
<tr>
<td>cartilage (M1, M2, M3)</td>
<td>41.9</td>
<td>0.83</td>
<td>1000</td>
</tr>
<tr>
<td>skin (M1, M2, M3, M4)</td>
<td>40.7</td>
<td>0.65</td>
<td>1100</td>
</tr>
<tr>
<td>fat (M1, M2, M3)</td>
<td>10.0</td>
<td>0.17</td>
<td>1100</td>
</tr>
<tr>
<td>muscle (M1, M2, M3, M4)</td>
<td>57.4</td>
<td>0.82</td>
<td>1040</td>
</tr>
<tr>
<td>brain, grey matter (M1, M2)</td>
<td>53.8</td>
<td>1.17</td>
<td>1030</td>
</tr>
<tr>
<td>brain, white matter (M1, M2)</td>
<td>34.5</td>
<td>0.59</td>
<td>1030</td>
</tr>
<tr>
<td>brain, average (M3, M4)</td>
<td>41.0</td>
<td>0.86</td>
<td>1030</td>
</tr>
<tr>
<td>CSF (M1, M2, M3)</td>
<td>79.1</td>
<td>2.14</td>
<td>1000</td>
</tr>
<tr>
<td>eye humour (M1, M2, M3, M4)</td>
<td>67.9</td>
<td>1.68</td>
<td>1010</td>
</tr>
<tr>
<td>eye lens nucleus (M1, M2, M3)</td>
<td>36.6</td>
<td>0.51</td>
<td>1050</td>
</tr>
<tr>
<td>eye lens outer (M1, M2, M3)</td>
<td>51.6</td>
<td>0.90</td>
<td>1050</td>
</tr>
<tr>
<td>eye sclera (M1, M2, M3)</td>
<td>54.9</td>
<td>1.17</td>
<td>1020</td>
</tr>
<tr>
<td>blood (M3)</td>
<td>55.0</td>
<td>1.86</td>
<td>n/a</td>
</tr>
<tr>
<td>parotid gland (M3)</td>
<td>70.0</td>
<td>1.90</td>
<td>n/a</td>
</tr>
</tbody>
</table>

Table 3.3: Tissue parameters for the experimental simulations

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$\varepsilon_r$</th>
<th>$\sigma$ [mho/m]</th>
<th>$\rho$ [kg/m$^3$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>bone (E1, M4)</td>
<td>14.8</td>
<td>0.15</td>
<td>1850</td>
</tr>
<tr>
<td>skin (E1, M4)</td>
<td>42.0</td>
<td>0.78</td>
<td>1100</td>
</tr>
<tr>
<td>muscle (E1, M4)</td>
<td>51.7</td>
<td>1.11</td>
<td>1040</td>
</tr>
<tr>
<td>brain, average (E1, M4, E2, E3)</td>
<td>41.0</td>
<td>0.88</td>
<td>1030</td>
</tr>
<tr>
<td>eye</td>
<td>67.9</td>
<td>1.68</td>
<td>1030</td>
</tr>
</tbody>
</table>

Integration Technique (FIT), which is well described in [17]. This technique is conceptually slightly different than FDTD but leads to the same numerical scheme. The open domain is bounded by second-order Mur absorbing boundary conditions. Excitation is done by a smoothly-increasing harmonic function and the computation is terminated after steady state is reached (usually after 10 to 20 periods). Computational time for 2.5 Million voxels was typically 3 hours on an IBM RS6000/560 computer.
3.2.2 Experimental Technique

The setup shown in Figure 3.1 was used to perform the experimental studies. The SAR distribution is determined by measuring the electric field with miniaturized E-Field probes inside shell phantoms filled with tissue simulating liquids. The probes are positioned by a 6-axis precision robot (Stäubli RX90) with a position repeatability of better than ±0.02 mm (at constant temperature). An optical surface-detecting system is integrated into the probes, which enables the accurate positioning of the probe with respect to the phantom's inner surface. The SAR distribution can be mea-
Figure 3.3: Numerical phantom which corresponds to the experimental head phantom $E_1$.

...ured for basically any volume. For dosimetric assessments of RF-sources close to the head, the following measurement strategy has been implemented: The E-field probe first scans over a large area inside the head to roughly localize the maximum SAR value. In a subsequent step SAR-measurements are done along a fine grid within a 35 g volume cubically shaped around this maximum value. This cube is large enough to provide enough data to evaluate the spatial peak SAR. The whole procedure is completely automated and takes less then 15 minutes. A detailed description of the system is given in [4]. An error analysis has shown that the uncertainty of the measured spatial peak SAR values is less than 20%, using the probe calibration technique as described in Chapter 2.

3.3 Head Phantoms

3.3.1 Exposure

In order to avoid the uncertainties of modeling actual handsets, a dipole $0.45\lambda$ in length was chosen. It was positioned at a distance of 15 mm from the head and with an orientation parallel to the body’s axis (see Figures).

In the numerical approach the dipole was simulated as a filament $0.45\lambda$ in length. A SPEAG 900 MHz test dipole of exactly the same length and made of standard 3.6 mm semi-rigid coaxial cable was used for the measurements. The bifilar matching line had a length of exactly $\lambda/4$ at 900 MHz, allowing direct measurement of the feedpoint impedance of the dipole, using a network analyzer with a shorted feeding gap as reference. Accurate information about the feed point impedance which changes considerably with distance from the body is essential, since the absorption is not primarily proportional to the output power but to the square of the antenna current [5]. In order to compare the results, all values were therefore normalized to an antenna current of 100 mA.

3.3.2 Numerical Phantoms

Accurate phantoms of heterogeneous human heads can be generated on the basis of magnetic resonance imaging (MRI). The translation of the three-dimensional data sets of relaxation times into the tissue distribution is a difficult task and generally
Figure 3.4: Photo of the experimental head phantom El. 5 tissues are simulated: brain, bone, skull, muscle and skin.

requires a person trained either in medicine or biology, who is able to distinguish both transitional and marginal regions. MRI produced in different laboratories, by different scientists and from different test subjects inevitably contain differing discretizations.

In this study three phantoms discretized by different groups, based on MRI data sets of three different adults, are used and their results compared. Figure 3.2 shows the outer shapes and two cross-sectional views of these head phantoms and Table 3.1 gives the data of discretization. The ear closest to the antenna needs special attention:

During normal use of a hand-held telephone, the ear is pressed against the head and, therefore, changes its shape. In order to avoid effects caused by the different ear modeling, which may mask the effects of the head itself, the outer right ear of the head phantoms was removed.

The phantoms can be described as follows:

- Model $M1$ was taken with the highest resolution. Its voxel size is $1\,\text{mm}$ in all three cartesian dimensions. $M1$ has the largest volume. The brain region was segmented very carefully. For the entire head 13 tissue types were simulated. However, the lower part of the head was assigned to only one tissue type.

- The second head phantom $M2$ has nearly the same voxel size. It was developed for the training of medical students and distinguishes among 120 tissue types. For the EM analysis this large number needed to be reduced to 13 different
tissues for which electric parameters are available [9].

- The third head phantom M3 was taken with a voxel size of about 12 mm$^3$. The discretization is relatively crude. In the original MRI model the skin was not identified. In the computer model the skin was added as an outer layer with a thickness of one voxel. The brain region of head phantom M3 is homogeneous and assigned only one tissue type.

- The fourth numerical phantom M4 (see Figure 3.3) corresponds to the experimental phantom El described below (Figure 3.4). It was derived from MR and CT cross sections taken every 2 mm. The voxel size was 2 mm $\times$ 2 mm $\times$ 2 mm.

Comparing the data of different publications reveals a spread in the values given for the electric parameters of different types of tissue. Table 3.2 shows the electric parameters we have chosen for this study. The permittivity and the conductivity of the tissues of phantoms were taken from the dielectric database [9]. Since M3 did not distinguish between different brain tissues, an averaged value of white and gray matter for brain tissue was used. In order to better compare the results from the numerical phantoms with those from the experimental phantoms, M4 was simulated with the tissue properties of Table 3.2, as well as with those of Table 3.3.

With numerical simulations, using FIT or other finite-difference codes, it is easy to attribute different tissue parameters to different mesh cells. However, the tissue discretization and the assignment of the electrical parameters to various tissues is
fraught with considerable uncertainties. Therefore, the question of whether these parameters significantly alter the absorption has been studied by relating different parameters to the various tissues:

- anatomically correct head phantoms with tissue distributions derived from MRI (referred to as M1, M2, M3, M4);
- simplified head phantoms, which have the outer shape of the MRI phantoms, but which contain only one tissue with high water content ($\varepsilon_r = 43.5$, $\sigma = 0.9 \text{ mho/m}$) and one with low water content, using the parameter of bone tissue, i.e., $\varepsilon_r = 21$, $\sigma = 0.33 \text{ mho/m}$ (referred to as M1s, M2s, M3s, M4s); and
- homogeneous head phantoms with the outer shape of the MRI phantoms which contain only one tissue type with $\varepsilon_r = 43.5$, $\sigma = 0.9 \text{ mho/m}$ (referred to as M1h, M2h, M3h, M4h).

### 3.3.3 Experimental Phantoms

Three different phantoms were studied.

- *Complex, Five Tissue Model* referred to as E1: artificial “*in-vivo*” phantom head from Microwave Consultants (Figure 3.4). The phantom simulates four tissue types: skull, muscle, skin and eyes. The fifth tissue, brain, is simulated by a sugar-water-salt solution with the electric parameters of the mean value
Figure 3.7: SAR distribution in the xz-plane of head phantom M1 (left), M2 (middle) and M3 (right).

Figure 3.8: SAR distribution in the xy-plane of head phantom M1 (left), M2 (middle) and M3 (right).

Figure 3.9: SAR distribution in the xx-plane of head phantom M1s (left), M2s (middle) and M3s (right).

Figure 3.10: SAR distribution in the xz-plane of head phantom M1h (left), M2h (middle) and M3h (right).
between gray and white matter \((\epsilon_r = 41, \sigma = 0.88 \text{ mho/m})\). The parameters of the other tissues are given in Table 3.3. The head size does not correspond to that of an average person since the volume of the brain is only about 1 liter instead of the typical value amounting to about 1.4 to 1.5 liters. In addition the skin-skull layer in the area above the ear is wider (15.5 mm) than for an average person.

- **Homogeneous Shell Copy** referred to as \(E_2\): exact copy of the outer shape of the previous complex phantom (Figure 3.5). The shell is made of polyester \((\epsilon_r = 4)\) and the thickness varies between 3 mm and 6 mm due to the manufacturing process. This phantom is again filled with the same brain tissue simulating liquid. About 2.5 liters can be used. In this case not only the brain area but the entire human head is modeled homogeneously.

- **Homogeneous Torso Model** referred to as \(E_3\): head-torso phantom of a human being (Figure 3.6). The shell is anatomically correctly shaped and is made of a 3 mm thick fiberglass \((\epsilon_r = 2.8)\). Shape and size of this phantom are very different from the shell copy of the complex phantom. It is considerably larger (4 liters instead of 2.5). For dosimetric characterizations of MTE this phantom is currently used in several laboratories in Europe and the United States.

### 3.4 Results

#### 3.4.1 SAR Distribution in Various Head Phantoms

The coordinate system used for analysis and presentation of the results is oriented so that the \(z\)-axis is parallel to the head-axis and the dipole-axis. The \(x\)-axis runs from the body-axis towards the dipole feedpoint. The dipole feedpoint is always located at \(x = 115 \text{ mm}, \ y = 0, \ z = 0\).

Figure 3.7 and Figure 3.8 show the qualitative SAR distribution of each of the anatomically correct head phantoms in both the \(xz\)-plane and the \(xy\)-plane respectively. The dipole is seen as a line or a point. High SAR values are in red, low SAR values in blue. The results can be summarized as follows:

- The region with high absorption values in all head phantoms is small and close to the feedpoint of the dipole. In most parts of the head the EM field is relatively low.

- Two SAR maxima can be identified in each of the MRI based phantoms, one on the skin's surface and one on the brain's surface. The SAR induced in bone tissue is considerably lower. The absolute values of the maxima differ from model to model.

- Examining the details of the SAR distribution in the brain region, one can identify varying SAR distributions in phantoms M1 and M2, which may be attributed to different conductivities of gray and white matter. Phantom M3 reveals a monotonic decay of the absorption rate inside the brain tissue.

Figure 3.9 depicts the simplified head phantoms with only two types of tissue. The resulting SAR distribution shows some agreement and differences with respect to that of the anatomically correct phantoms:
The simplified phantoms also exhibit the two SAR maxima at the skin surface and the brain surface with the minimum in the bone in between. The maxima at the brain surface in Figure 3.9 are lower than those in Figure 3.7. This is primarily due to the higher conductivity ($\sigma = 0.9 \text{ mho/m}$) which is used for the skin in the simplified phantoms in comparison to the actual skin conductivity of 0.65 mho/m.

Since no tissues are distinguished inside the brain region, the SAR distribution shows a smooth decay.

Figure 3.10 shows the SAR distributions inside the homogeneous head phantoms M1h, M2h and M3h. These differ from those of the anatomically correct and simplified phantoms as follows:

- Only one SAR maximum is observed at the phantom surface.
- Inside the phantom the EM field decays monotonically.

### 3.4.2 Averaged SAR Values

All these differences can be directly explained by the energy absorption mechanism. According to [5], the induced SAR is primarily determined by the square of the H-field which drops inversely proportional to the square of the distance from the
Figure 3.12: SAR profile in the simplified head models (M1, M2, M3, M4) and in one experimental model (E1) at 900 MHz above the antenna feedpoint. Water-containing tissues were substituted by brain tissue.

<table>
<thead>
<tr>
<th>Phantom</th>
<th>$I_{rms}$ [mA]</th>
<th>$P_{in}$ [W]</th>
<th>$P_a/P_{in}$</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>anatom. corr.</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M1</td>
<td>100</td>
<td>0.44</td>
<td>0.72</td>
</tr>
<tr>
<td>M2</td>
<td>100</td>
<td>0.48</td>
<td>0.69</td>
</tr>
<tr>
<td>M3</td>
<td>100</td>
<td>0.52</td>
<td>0.69</td>
</tr>
<tr>
<td><strong>simplified</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M1s</td>
<td>100</td>
<td>0.43</td>
<td>0.74</td>
</tr>
<tr>
<td>M2s</td>
<td>100</td>
<td>0.46</td>
<td>0.69</td>
</tr>
<tr>
<td>M3s</td>
<td>100</td>
<td>0.51</td>
<td>0.70</td>
</tr>
<tr>
<td><strong>homogeneous</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M1h</td>
<td>100</td>
<td>0.41</td>
<td>0.72</td>
</tr>
<tr>
<td>M2h</td>
<td>100</td>
<td>0.41</td>
<td>0.65</td>
</tr>
<tr>
<td>M3h</td>
<td>100</td>
<td>0.41</td>
<td>0.70</td>
</tr>
<tr>
<td>E2</td>
<td>100</td>
<td>0.45</td>
<td>-</td>
</tr>
<tr>
<td>E3</td>
<td>100</td>
<td>0.46</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 3.4: Antenna current, antenna input power and ratio of absorbed power to input power of various head phantoms
Figure 3.13: SAR profile in the homogeneous head models (M1, M2, M3, M4), the sphere and the experimental models (E1, E2, E3) at 900 MHz above the antenna feedpoint.

source. In the case of transmitters very close to the head, this factor is dominant so that even lossy bone tissue can be approximated by a layer of air of the same thickness in order to assess the SAR in the brain tissue behind this layer. In other words, the distance dependence dominates the field attenuation due to absorption. The often-discussed effects occurring in far field exposure situations (e.g., refraction effects, etc.) are negligible in case of the extreme nearfield exposure. Quantitative presentations of the SAR values in Figures 3.11 to 3.13 on the x-axis in the range 40 mm ≤ x ≤ 100 mm confirm this. The head surface facing the dipole is located at x = 100 mm. In addition, in the same figures the simulations are compared with the measurement data which are normalized to an antenna feedpoint current of 100 mA. Noticeable is the good agreement between measured and simulated data.

The volume-averaged spatial peak SAR values (mW/cm³) were derived by shifting a cube of 10 mm (1 g) or 21.5 mm side length (10 g) over the head region, calculating $\bar{\sigma}|E|^2$ averaged over each cube and searching for the position where this value is at a maximum.

The averaged values for the experimental phantoms are derived as described earlier. 175 measurements inside a cube of about 35 g of brain tissue provided sufficient data to accurately locate the spatial peak SAR values averaged over 1 g and 10 g using interpolation and extrapolation routines.

This is a departure from the definition in the standard requiring averaging over a mass of 1 g or 10 g tissue, but is much easier to compute and to compare. It, however,
slightly underestimates the mass-averaged SAR values.

In Figures 3.14 and 3.15 SAR values averaged over $1\,\text{cm}^3$ and $10\,\text{cm}^3$ for the various heads and different modelings are compared. The difference between the various heads and the different modelings are surprisingly small. The smaller values of phantom M3 can be explained by the thickness of the bone layer which exceeds that of an average adult. The homogeneous head overestimates the absorption compared to the highest value found among the inhomogeneous phantoms by 25% for the $1\,\text{cm}^3$-average and by only 10% for the $10\,\text{cm}^3$-average.

In Table 3.4 the antenna input power $P_{\text{in}}$ as well as the ratio of absorbed $P_a$ to the antenna input power are given. The radiated power is about $450\,\text{mW}$ and varies by 20% depending on the modeling. Typically, 70% of the antenna input power is absorbed by the head. The variation of $<15\%$ indicates that homogeneous heads are suitable for the purpose of radiation optimization of MTE.
3.5 Conclusion

The results basically confirm the conclusions of [5] that the spatial peak SAR is scarcely affected by the size and the shape of the human head for electromagnetic sources at a defined distance from the human head. Compared to other factors, such as distance of the source from the head and design of the devices, the effects caused by the complex anatomy are minor especially in the case of volume-averaged values. Due to the strong radial decay of the H-field in proximity to the source, variations of the bone conductivity have scarcely any effect on the absorption in the brain tissue.

The comparison of the results obtained from the inhomogeneous and homogeneous phantoms suggests that homogeneous phantoms are highly suited to be used in compliance tests for handheld MTE operating in the 900 MHz band. The overestimation of the averaged spatial peak SAR values is small when compared to the largest value obtained by the inhomogeneous phantoms. This is especially true if the values are averaged over a volume equivalent to 10g. The major advantage of using simple homogeneous phantoms is that the number of tests can be reduced since small shifts of the source parallel to the surface result in almost no changes in the spatial peak SAR values. In case of inhomogeneous modeling, variations of several dB must be expected for shifts of a few millimeters. Therefore, inhomogeneous phantoms might result in artificially low SAR values for some positions of the device with respect to the head, which would not represent the actual exposure of the various users.

These results were found for sources operating at 900 MHz but can be extended for frequencies down to 300 MHz, due to the absorption mechanism. Whether they are valid for the 1.5 to 2 GHz band must be the subject of further study.
References


Chapter 4

The Dependence of EM Energy Absorption upon Human Head Modeling at 1800 MHz

Abstract - The previous chapter examined the dependence of electromagnetic energy absorption on human head modeling for transmitters operating at 900 MHz in close proximity to the head. The same question is now extended to the frequency bands utilized by the new mobile communications systems between 1.5 GHz and 2.5 GHz. Additional studies became necessary since some peripheral tissue layers reach a thickness in the range of \( \lambda/4 \) to \( \lambda/2 \). The conducted simulations and worst case considerations confirmed the expected, more complex relation between absorption and anatomical details at these higher frequencies. Nevertheless, a homogeneous representation of the head is suited for assessing the maximum SAR in the head of the user of an MTE if the appropriate dielectric parameters are chosen.

4.1 Introduction

Due to an increased demand for bandwidth, new generation Mobile Telecommunications Equipment (MTE) are shifting towards higher frequency bands. PCN or PCS systems operate within the frequency band of 1.5 GHz to 2.5 GHz. In the case of handheld MTE, the transmitter operates in immediate proximity to the head. In 1994, the US Federal Communications Commission [1] requested demonstration of compliance of PCS systems with the ANSI/IEEE [2] standard for protection from non-ionizing radiation. Recently the International Commission on Non-Ionizing Radiation Protection (ICNIRP) [3] called for the compliance of all MTE with the basic restrictions of the ICNIRP Standard. A working group of CENELEC is currently in the process of developing such a testing standard for MTE based on a mandate of the European Community [4], [5].

Two dosimetric E-field scanning systems have been developed for such compliance tests, both of which are restricted to measurements in shell phantoms filled with a tissue-simulating liquid [6], [7]. The suitability of such homogeneous head phantoms...
has been investigated in the previous chapter on the dependence of electromagnetic (EM) energy absorption upon human head modeling at 900 MHz. The results showed that, by using homogeneous head phantoms, the assessed spatial peak SAR is always larger than the maximum value occurring among all users. Most important is the finding that the overestimation for the region above the ear is modest, i.e., less than 10% for the an averaging mass of 10 g and less than 25% for 1 g. On the basis of the energy absorption mechanism [8], these findings can be extended to frequencies as low as 300 MHz. An extension to higher frequencies was not possible without further studies since peripheral tissue layers are expected to play a more prominent role than at lower frequencies. The reasons are that the thickness of some tissue layers are in the range of λ/4 to λ/2 whereby the attenuation in these layers is not significant enough to exclude possible enhancement effects due to reflection at the boundaries or due to matching effects.

The objective of this chapter was to assess these effects with respect to the suitability of homogeneous phantoms for testing compliance of MTE operating at 1800 MHz. The methods applied and the approaches taken were basically identical to those for the study at 900 MHz in Chapter 3. The spatial peak SAR averaged over 1 g and 10 g tissue mass were compared for different non-homogeneous and homogeneous phantoms of the human head. In addition, the findings were strengthened with "worst case" considerations.

4.2 Methods

4.2.1 Numerical and Experimental Techniques

For the numerical analysis of the complex head phantoms the software package “MAFIA”, based on the finite integration technique (FIT), was used. This technique, although slightly different from the FDTD technique conceptually, nevertheless results in the same numerical scheme. The open domains are bounded by second-order Mur absorbing boundary conditions. Details are given in [9]. Excitation was done by a smoothly-increasing harmonic function and the computation was terminated after steady state was reached (usually after 10 to 20 periods). Computational time for 2.5 Million voxels was typically 3 hours on an IBM RS6000/560 computer.

For the spherical phantoms the 3D MMP software package was applied. This frequency domain boundary technique enables reliable error estimations and is especially efficient and accurate for layered spheres. Details are given in [10].

The dosimetric assessment system DASY2 was used to experimentally assess the absorbed electromagnetic power inside the different shell phantoms. The uncertainty for the spatial peak SAR is given to within ±10% and the repeatability lies within ±5% [7]. The system and its calibration procedure are described in detail in [7] and Chapter 2.

4.2.2 Head Phantoms

The four different complex numerical head phantoms M1 to M4 are described in chapter 3. M1 to M3 are based on MRI scans of three different adults, each varying considerably in discretization, size and shape. The fourth model M4 corresponds to the most complex experimental model E1 which was discretized based on MRI- and CT-scans. The tissue parameters assigned to these phantoms are summarized
Table 4.1: Tissue parameters of the numerical models at 1800MHz

<table>
<thead>
<tr>
<th>Tissue</th>
<th>(\epsilon_r)</th>
<th>(\sigma)</th>
<th>(\rho)</th>
<th>(\lambda)</th>
</tr>
</thead>
<tbody>
<tr>
<td>bone (M1, M2, M3, M4)</td>
<td>19.7</td>
<td>0.55</td>
<td>1850</td>
<td>37</td>
</tr>
<tr>
<td>cartilage (M1, M2, M3)</td>
<td>38.3</td>
<td>1.23</td>
<td>1000</td>
<td>27</td>
</tr>
<tr>
<td>skin (M1, M2, M3, M4)</td>
<td>38.4</td>
<td>0.99</td>
<td>1100</td>
<td>27</td>
</tr>
<tr>
<td>fat (M1, M2, M3)</td>
<td>9.4</td>
<td>0.25</td>
<td>1100</td>
<td>54</td>
</tr>
<tr>
<td>muscle (M1, M2, M3, M4)</td>
<td>55.2</td>
<td>1.30</td>
<td>1040</td>
<td>22</td>
</tr>
<tr>
<td>brain, grey matter (M1, M2)</td>
<td>51.5</td>
<td>1.57</td>
<td>1030</td>
<td>23</td>
</tr>
<tr>
<td>brain, white matter (M1, M2)</td>
<td>33.1</td>
<td>0.84</td>
<td>1030</td>
<td>29</td>
</tr>
<tr>
<td>brain, ref. experiments (M3, M4)</td>
<td>41.0</td>
<td>1.65</td>
<td>1030</td>
<td>26</td>
</tr>
<tr>
<td>CSF (M1, M2, M3)</td>
<td>77.8</td>
<td>2.83</td>
<td>1000</td>
<td>19</td>
</tr>
<tr>
<td>eye humour (M1, M2, M3, M4)</td>
<td>67.2</td>
<td>2.08</td>
<td>1010</td>
<td>20</td>
</tr>
<tr>
<td>eye lens nucleus (M1, M2, M3)</td>
<td>34.7</td>
<td>0.87</td>
<td>1050</td>
<td>-</td>
</tr>
<tr>
<td>eye lens outer (M1, M2, M3)</td>
<td>49.5</td>
<td>1.33</td>
<td>1050</td>
<td>-</td>
</tr>
<tr>
<td>eye sclera (M1, M2, M3)</td>
<td>52.7</td>
<td>1.67</td>
<td>1020</td>
<td>-</td>
</tr>
<tr>
<td>blood (M3)</td>
<td>54.0</td>
<td>2.27</td>
<td>n/a</td>
<td>22</td>
</tr>
<tr>
<td>parotid gland (M3)</td>
<td>70.0</td>
<td>1.90</td>
<td>n/a</td>
<td>20</td>
</tr>
</tbody>
</table>

Table 4.2: Tissue parameters of the experimental models at 1800MHz

<table>
<thead>
<tr>
<th>Tissue</th>
<th>(\epsilon_r)</th>
<th>(\sigma)</th>
<th>(\rho)</th>
<th>(\lambda)</th>
</tr>
</thead>
<tbody>
<tr>
<td>bone (E1)</td>
<td>13.4</td>
<td>0.30</td>
<td>1850</td>
<td>45</td>
</tr>
<tr>
<td>skin (E1)</td>
<td>32.5</td>
<td>1.98</td>
<td>1100</td>
<td>28</td>
</tr>
<tr>
<td>muscle (E1)</td>
<td>50.6</td>
<td>1.56</td>
<td>1040</td>
<td>23</td>
</tr>
<tr>
<td>brain, experimental (E1, E2, E3)</td>
<td>41.0</td>
<td>1.65</td>
<td>1030</td>
<td>26</td>
</tr>
<tr>
<td>eye</td>
<td>67.6</td>
<td>2.26</td>
<td>1030</td>
<td>20</td>
</tr>
</tbody>
</table>

in Table 4.1. In addition to the non-homogeneous modeling, the internal tissue distribution was simplified. In the first step, the parameters of average brain tissue were assigned to all tissues except the bone structure which was simulated as bone tissue. These phantoms are referred as simplified phantoms in the following. Finally the bone tissue was replaced by brain-average tissue (referred as homogeneous phantoms).

The three experimental phantoms are also described in detail in Chapter 3. The experimental phantom E1 is commercially available and simulates five tissue types: skin, muscle, bone, eyes and brain. Brain is simulated by a liquid. The second model
Figure 4.1: SAR profile in the complex head models (M1, M2, M3, M4) at 1.8 GHz above the antenna feedpoint.

Figure 4.2: SAR distribution in the xz-plane of head phantom M1 (left), M2 (middle) and M3 (right).

E2 is a polyester shell copy of E1 filled with a simulating liquid. The third model E3 is a fiberglass shell phantom currently used in various laboratories in the USA and Europe for compliance tests.

In order to avoid the uncertainties of modeling actual handsets, a dipole 0.45λ in length was chosen. It was positioned at a distance of 15 mm from the head and oriented parallel to the body's axis as shown in Figure 4.2. In the numerical approach the dipole was simulated as a filament 0.45λ in length. A SPEAG 1800 MHz test
Figure 4.3: SAR profile in the simplified head models (M1, M2, M3, M4) at 1.8 GHz above the antenna feedpoint. Water-containing tissues were substituted with brain tissue.

dipole of exactly the same length and made of standard 2.2 mm semi-rigid coaxial cable was used for the measurements. The dipole is constructed the same way as described in Chapter 3 enabling accurate measurement of the feedpoint impedance. All results shown in the following were therefore normalized to an antenna feedpoint current of 100 mA.

4.3 Results and Discussion

Figure 4.2 shows the SAR distribution in the cross section of the three numerical head phantoms M1 - M3 whereas Figures 4.1 to 4.4 summarize the SAR values on the x-axis. The x-axis runs through the body's axis as well as through the feedpoint of the dipole located at $x = 115$ mm and oriented in z-direction.

At 1800 MHz most of the induced energy is already absorbed in the peripheral skin-muscle layer compared to 900 MHz because the penetration depth is reduced by 50%; i.e., the second maximum at the brain layer is much less pronounced than at lower frequencies. The values at the surface are nearly twice as large as those at 900 MHz which corresponds well with the approximation formula given in [8]. The determining factors are the larger conductivity and, more importantly, the increased coupling due to the larger distance of the dipole in terms of wavelength (i.e. larger value of the correction factor in [8]).

The SAR distribution through the layers deviates from the proportionality of
Figure 4.4: SAR profile in the homogeneous head models (M1, M2, M3, M4), the sphere and the experimental models (E1, E2, E3) at 1.8 GHz above the antenna feedpoint.

The matching effect between free space and brain tissue is expected, if water-containing tissue is covered by a layer of intermediate dielectric parameters and layer thickness of approximately λ/4. That such effects can also occur for nearfield exposure is shown by the simulations for a spherical brain phantom covered by a bone shell of 10 mm thickness (Figure 4.7).

Any differences in the SAR distribution disappear as expected in the case of homogeneous modeling of the various phantoms. The value at the surface also corresponds well with the 17 mW/g estimated by the approximation formula in [8].

The volume-averaged SAR values are compared in Figures 4.5 and 4.6. The 10 cm$^3$-averaged values are about 5% higher than those from the heterogeneous phantoms. In the case of the 1 cm$^3$ averaging volume, the homogeneous numerical results as well as the experimental results are 20% higher. The main reason for these rather large differences lies in the significant differences in conductivity between the parameters of skin where most of the energy is absorbed and those used in the simplified and homogeneous phantoms. The parameters corresponded to the experimental
brain tissue liquid (Table 4.1).

Regarding compliance tests it is important to note that specific tissue layers in the range of the head’s anatomy may not only increase the local SAR values but also the spatial peak SAR values (Figure 4.5 and 4.6). Based on a simple layered half space exposed to a plane-wave, an estimate for the maximum enhancement of the spatial peak SAR values compared to the homogeneous half space ($\epsilon_r = 41, \sigma = 1.65$ mho/m) had been analytically studied by varying the thickness of the front layers. The results are summarized in Table 4.3 for different layered structures: skin-muscle, skin-bone, muscle-bone and bone-brain. The parameters of Table 4.1 were used.

The enhancement effects observed in the case of the structure bone-brain vanish if the bone layer is covered by a skin layer of thicker than 1 mm. It can be concluded that a homogeneous phantom consisting of material with the parameters ($\epsilon_r = 41,$
Figure 4.7: Influence of a peripheral bone and skin layers on the absorption. Computed with layered spherical models at 1.8 GHz. Dipol distance was 15 mm.

<table>
<thead>
<tr>
<th>Structure layer</th>
<th>SAR(_{10g-avg})</th>
<th>SAR(_{10g-avg})</th>
</tr>
</thead>
<tbody>
<tr>
<td>skin</td>
<td>muscle</td>
<td>0.87</td>
</tr>
<tr>
<td>skin</td>
<td>bone</td>
<td>0.85</td>
</tr>
<tr>
<td>muscle</td>
<td>bone</td>
<td>0.87</td>
</tr>
<tr>
<td>bone</td>
<td>brain</td>
<td>1.1</td>
</tr>
</tbody>
</table>

Table 4.3: Maximum enhancement factors (EF) of layered structures compared to a homogeneous half space (\(\varepsilon_r = 41, \sigma = 1.65\) mho/m) under plane wave exposure (1800 MHz)

\(^a\)Maximum enhancement is found with no front layer due to the lower reflection of bone tissue and the bigger averaging volume.

\(\sigma = 1.65\) mho/m) overestimate the actual exposure for all users. This is especially true if the distance dependence of the incident H-field is considered.

4.4 Conclusions

In the frequency band of PCS and PCN anatomical layered structures of the human head can lead to increased absorption in the layers. However, the spatial peak SAR assessed by a homogeneous phantom with the parameters of grey brain tissue are unlikely to be less than the actual SAR value induced among all potential users.
and operational conditions, i.e., it would satisfy the worst case criteria of [4]. The overestimation is less than 5% for the 10g-averaged spatial peak SAR and significantly less than 20% for the 1g-averaged value.
References


Leer - Vide - Empty
Part IV

Homogeneous Modeling For Compliance Tests
Leer - Vide - Empty
Chapter 5
Consideration of Additional Factors

Abstract- In the two previous chapters the influence of anatomy and modeling on the absorption of electromagnetic energy in the human head in the frequency range of mobile communications was investigated for RF-sources in very close proximity to the head in a "worst case" exposure situation. In the following chapter the question is addressed, whether such an approach based on a homogeneous model incorporates the influence of strongly heterogeneous tissue distributions in the ear region, the exposition of children using MTEs and additional factors which are not directly tied to the anatomy of the head but which also influence the induced SAR, such as the hand holding the device and external and internal metallic structures in the head region, such as glasses and implants.

5.1 Introduction

In the first chapter various complex and interdependent factors were discussed which determine the induced SAR in the head of an MTE user:

- Relative position of the MTE with respect to the head. There is considerable variation in the operating positions used in operating a mobile telephone, and the telephone can be held in various ways.

- Size and shape of the head and the complex heterogeneous anatomy in the head. The shape of the human head shows strong individual differences.

- Variations in the manufacturing of the device, the specified output power, the antenna design, the mounting of the antenna on the device (left/right) and the antenna operating position (in/out).

- External and internal metallic objects close to the head or in the head which are excited by the MTE, behaving as passively radiating structures.
CENELEC's WGMTE preliminary draft prENV 501066-2 [1] for standardized compliance testing requires that all users be covered by the assessment method. In addition, all assessment-uncertainties must be added to the measured value.

A standardized model for the human head should, therefore, enable coverage of the above-mentioned factors and guarantee that the assessed SAR values are never smaller than in reality, adding - as required in the WGMTE proposal - all uncertainties to the measured value. The complex interdependencies of the device and the human body must be taken into account such that real operating conditions can be recreated and tested.

The dependence of electromagnetic energy absorption upon human head modeling was investigated in the two previous chapters. The results demonstrated that anatomical inhomogeneities only very slightly influence the volume-averaged SAR and that size and shape have barely any influence on the SAR. However, in those studies a "worst case" exposure situation was chosen. The dipole exposed a region of the head above the ear where a simple anatomy is encountered with a thin skin layer followed by thin muscle and bone layers before the brain is reached. Based on the results, it was concluded that a homogeneous modeling of the human body is suitable to describe the "worst case" exposure of the user of an MTE regarding today's safety limits.

In this chapter this "worst case" approach to operating conditions is expanded. Since a mobile telephone can be held in various operating positions, other more complex and heterogeneous regions of the head, such as around the ear, can be exposed. The induced SAR values will strongly depend on the relative position of the source next to such complex structures. Consequently, small shifts of the source will lead to large variations in the averaged SAR. The sensitivity of the SAR to such shifts in the anatomically highly heterogeneous region and the probable overestimation when homogeneously modeling the head had to be determined.

Further, since only adult heads had been investigated thus far, the question whether children were also covered by such a homogeneous model due to their differing anatomies, had to be investigated as well.

In the last section of this chapter the possible influence of additional factors not directly tied to anatomical questions, are addressed: the role of the hand holding the device, metallic structures at the head surface, such as glasses and jewelry, and internal metallic structures, such as head and dental implants. If it is not feasible to take these factors into account with the design of the phantom, then the restrictions must be amended.

5.2 Variations in the Ear Region

The investigations regarding the SAR dependence upon anatomical complexities were performed in a head region above the ear, where thin skin, muscle and bone tissue layers are encountered in front of the brain. The brain is a bulky and highly water containing body and, due to its physiological importance, was the main object of concern. Thus, this head region was seen as "worst case" prone. In addition, the tissue distribution is only inhomogeneous from the outside to the inside (i.e., along an axis normal to the surface). When shifting the source along the head surface, no variations of the SAR were expected.

Around the ear the anatomical structure is more complex. A heterogeneous
variation of the tissue distribution is not only encountered normal to the head surface but also in tangential directions. A massive bone structure (petrous bone) reaches deep into the head and various air or liquid filled cavities exist behind the opening of the ear canal. A homogeneous representation of this area based on brain or skin tissue would overestimate the actual exposure in this region of tissues with low permittivity and low conductivity.

On the other hand, when slightly shifting the position of the source along the head surface, a heterogeneous representation of this area would lead to large variations in the SAR. Therefore, it is possible that a single assessment based on one definite position of the source could lead to a strong underestimation of the possible actual exposure.

5.2.1 SAR-Estimation in the Ear Region

An extensive, time consuming numerical analysis of the absorption in the ear region was bypassed by a simple estimation based on the absorption model of [2]. As in the studies of the previous chapters, the outer ear itself was again removed in the phantom. The goal was to determine the approximate variation of the volume-averaged SAR in the ear region when the position of the source is shifted in a vertical plane parallel to the ear 15 mm at its closest distance from the head surface.

To estimate these possible variations of the SAR in the ear region, the approach was based on the following simplifications:

- For every location in the head the absorption can be estimated by using the approximation formula in [2], assuming that the SAR is predominantly determined by the square of the magnetic field and, in particular, from the square of the distance of the source from the assessment site. The feasibility of this assumption at a frequency of 900 MHz was shown in Chapter 3, where the distance dependence (1/r²) was dominant in comparison to the attenuation in tissue.

- The magnetic field is assumed to be constant along the cross-section of the cube-shaped averaging volume.

Under these assumptions the local SAR can be determined by simply assigning a conductivity value to the “discretization” point in the head.

\[ SAR \sim \sigma H^2 \]  

Tests of this approach using the results found for the upper head region in Chapter 3 demonstrated that the averaged SAR can be sufficiently assessed to yield an estimate for the variations of the averaged local SAR.

The conductivity distribution was taken from cross-sectional photographic and MRI images (Visible Human Project). Six different tissue types were distinguished: skin, muscle, bone, CSF, grey and white matter [3]. The head was “discretized” in an area of 60 mm x 75 mm around the opening of the ear canal and 82 mm into the head. Each discretization point modeled a volume element with the corresponding conductivity value [3]. The size of one volume element was 7.5 mm x 5.0 mm x 1.5 mm. To cover lateral tissue variations, weighted tissue properties of neighboring volume elements were taken into account as well. The maximum averaged SAR
Figure 5.1: Estimation of the distribution of the maximum local averaged SAR-values (1 g) in the ear region when the source is shifted in a vertical plane parallel to the ear. Closest distance is 15 mm from the ear canal opening.

for one position of the dipole was then estimated by averaging the volume elements along a line through the center of these volume elements and running from the dipole feedpoint normal to the vertical plane into the head. This procedure was repeated for each of the 120 locations of the dipole in the vertical plane parallel to the ear.

Figure 5.1 shows the volume-averaged SAR estimation for an averaging volume of 1 cm$^3$. In this case the estimated averaged SAR values can vary up to 3 dB within a radius of only 10 mm around the ear canal opening. The minimum occurs in the opening of the ear canal. The maximum can always be found in the region 30 mm to 40 mm above the ear where the peripheral tissue layers are thin. Behind the ear the absorption values decrease strongly, due to the increased curvature of the head and thus the increased distance of the source from the body.

5.2.2 Conclusions for Compliance Tests

The conclusion that can be drawn for compliance tests from this simple estimation is that a heterogeneous representation of the human head can lead to strong under-estimation of the maximum SAR occurring in real life situations. Small shifts (10 mm) of the source along the head surface next to the complex structure of the ear can cause large variations in the averaged SAR (~3 dB). These variations could also explain the widely differing results published by different laboratories. Homogeneous modeling of the entire head, on the other hand, may considerably overestimate the maximum power induced in the head, if the maximum occurs in the ear region. However, it should also be noted that homogeneous models guarantee that the maximum is definitely covered by one single measurement and that the maximum SAR values were generally found in the more homogeneous regions behind or above the ear when
5.3 Absorption in Children

Thus far the absorption in adult heads has been investigated. Only in [4], was the initial data on absorption in the heads of 5 to 10 year old children presented. These numerical results show an increased absorption when exposed to a 835 MHz and 1900 MHz source (numerically modeled MTE) at a distance of about 14 mm from the ear.

Two main questions regarding the child's anatomy must be addressed beforehand when determining the absorption in the child's head. These questions have not been addressed in [4], where the adult geometry of the head was simply scaled to a child's head of the specific age:

- Differences in the electrical parameters of children's heads tissues in comparison to the adult. The amount of fatty tissue in the child's brain is proportionately smaller because myelination of the brain nerve cells is not yet complete, i.e., it contains proportionately more water. Higher water levels can also be found in child bone tissue, together with increased levels in the blood.

- Differences in the anatomical geometry of children's heads. This includes the shape and size of the head as well as the tissue layers' thicknesses.

These preparatory studies were based on the medical literature available [5], [6], [7],
Figure 5.3: Mean bone thickness (±95% confidence interval) dependent on the child's age. Based on measurements 40 mm above the ear (ref.[5] and [6]).

Figure 5.4: Median head circumference dependent on the child's age. Curve in the middle corresponds to the median, the upper curve to the 97th percentile, the lower to the 3rd percentile [8].
5.3.1 Child Brain Tissue Parameters

Considerable differences in the electrical parameters of a child are only observable during a period of 24 months after birth. Up until then the water content of brain tissue as well as the water and blood content of the inner bone tissues are higher than those of adults. A new-born child’s grey matter contains about 89% water and is reduced during growth to 82% for an adult [7]. White matter shows 87% water in a new-born child and 72% in an adult. Adult white matter contains more fat, i.e., myelin, than grey matter does. The strongest variations take place between birth and the first 12 months. From 12 months to 48 months the myelination of nerve cells decreases sharply until it is completed around the age of 16. Figure 5.2 shows an estimated extrapolation in the conductivity of brain tissues from the adult to the child based on this data.

5.3.2 Child Head Geometry

An overview of the skull bone thickness about 40 mm above the ear shows a nearly linear increase in the thickness from birth (1 mm) to an age of eight (5 mm) [5], [6].
After age eight the increase slows down until it reaches a "final" thickness at around the age of sixteen (7 mm).

Evaluating the head sizes (circumference, width and length) dependent on the child's age reveals that the strongest growth takes place again within the first 12 months after birth [8]. A newborn shows a head circumference of about 36±3 cm (median value). After 12 months it reaches approximately 46±3 cm and after 24 months already 50±3 cm. Adults have a head circumference of about 55±3 cm. Therefore, the only relevant difference between child and adult heads lies in the ratio between skull bone thickness and head circumference, which is considerably smaller for a child.

5.3.3 Conclusions for Compliance Tests

Beginning only 24 months after birth the electrical parameters of the child's head tissues hardly differ from those of an adult. The head size changes mostly within the first 12 to 24 months. Considerable differences between children's and adults' heads only exist regarding the bone layer thickness: a child has a thinner skull in relation to the skull size than an adult.

A heterogeneous modeling of the head based on adult data would lead to an underestimation of the exposure of a child due to the thinner low loss bone layer and the slightly higher conductivity values of children's (24 months) brain tissue. On the other hand, a homogeneous modeling in which the bone is omitted, results in slightly higher averaged SAR values, which would cover the electromagnetic energy absorption in the child head. Numerical computations on layered spheres at 900 MHz confirm this assumption (see Figure 5.5). A child's head was simulated with the electrical parameters of a 24 month old child and a circumference of 47 cm (r=75 mm) as a lower limit and 63 cm (r=100 mm) as a "worst case".

These investigations must be regarded as preemptive estimations about the exposure of children. Extensive numerical calculations are currently being performed based on MRI-data from three to six year old children to thoroughly clarify this issue. Nevertheless, there is no reason to assume different results.

5.4 Additional Non-Anatomical Factors

The influence of the anatomical complexities of the head was investigated and the data show that a homogeneous representation can cover all uncertainties regarding the worst case maximum averaged SAR. However, there are additional factors which, although not directly tied to the head anatomy, can also influence the induced SAR:

- The hand holding the device could cause increased SAR values in the head. A "tuning"-effect of the radiating structure (device plus antenna) by the hand and the lower arm cannot be excluded a priori.

- External metallic structures close to the body which are passively excited could cause increased SAR values, due to their proximity.

- Implanted metallic structures could pick up the induced field and lead to increased internal SAR values. Geometry effects due to edges and ridges could cause strongly increased peak levels.
With the exception of the hand holding the device, these questions have not been addressed thus far. Using both up-to-date numerical and experimental means, the uncertainties introduced by these factors were determined.

### 5.4.1 The Hand Holding the Device

The influence of the hand on the absorption was investigated regarding the design of the device and the way the device is held. The dosimetric scanner DASY2 was used, together with a liquid-filled shell phantom as a homogeneous representation of the human head and torso. The accuracy of the assessment of the volume-averaged SAR with DASY2 was within ±10%, the repeatability within ±5% [11]. A representative selection of 15 MTEs was tested operating at 450 MHz, 900 MHz and at 1.8 GHz. The devices operated at a defined up-link frequency and at the maximum specified power level. Each device was positioned once by a mounting device and several times by the hand of an actual person. When a person was holding the device, three holding positions were chosen. First, in the middle of the device, second at the top close to the antenna mounting site and finally at the bottom of the casing. The position of the device in respect to the head was similar to a CENELEC-proposed test position which yielded high absorption values. Close contact between the arm of the person holding the device and the head torso model was always maintained.

The dosimetric scanner first scanned for the location of the maximum value, since the hand could change the current distribution on the device and thus the location of the maximum SAR.

In Figure 5.6 the results are summarized. A decisive factor is the design of the device. The better the antenna is decoupled from the device casing, the less influence the hand has. In general, the highest volume-averaged SAR values were measured without a hand holding the device. Across the entire frequency range similar results
were found. When the antenna is touched, the SAR drops to 50% of the case when
no hand is holding the device. Holding the device in the middle or at the bottom of
the device decreases the SAR by about 10%.

In conclusion, the hand does not lead to higher induced SAR values which cannot
be covered by a homogeneous model. Though the SAR values can drop considerably
when a hand is involved, a device can always be held in a way which scarcely affects
the SAR value. Therefore, the maximum volume averaged SAR can best be assessed
when no hand is simulated holding the device. The error will be within ±5%. Fi¬
nally, the hand can change the current distribution on the device. A heterogeneous
modeling of the head would mean that, especially in the anatomically complex ear
region, the modeling of the hand would become a decisive factor. A homogeneous
modeling, however, does not require the modeling of the hand.

5.4.2 External Metallic Structures

Metallic structures such as glasses can become passively radiating structures in close
proximity to the body excited by an MTE. Due to the closeness of this structure to
the body, higher SAR values would be induced. These effects were investigated at
900 MHz. For higher frequencies the effects were expected to be smaller, due to the
higher damping of the passive element by the lossy body.

To investigate the possible increase in SAR values from external metallic struc¬
tures, the dosimetric nearfield scanner DASY2 was used [11]. A “worst case” setup
was chosen: a resonant metallic thin rod was placed adjacent to the head, i.e., it
was attached to the shell phantom and the induced SAR was measured inside the
simulating liquid. Different arrangements of the relative mutual position of the pas¬
active and active structure were tested and the one with best coupling between the
structures was chosen. This is the case when the two elements are parallelly oriented
and the active element is in free space, i.e., not also adjacent to the body. The reso¬
nant length was first determined for a defined distance (15 mm) in air of the active
(test dipole) from the passive element. Due to the damping effects of the body on
the passive structure, the resonance length differs considerably from the free space
length (Figures 5.7 and 5.8).

A resonant structure adjacent to the head causes increased induced SAR values
in the head due to its proximity compared with the case where no passive structure
is present (Figure 5.8). However, the values are always smaller than when the active
structure touches the body (Figures 5.7 and 5.9).

The passive element shows strong influence on the feedpoint impedance of the
antenna, depending on the distance between passive and active elements (Figure
5.10). It is theoretically possible that, by adaptive compensation of the reflected
power, excessive SAR values could occur. Thus, the measurements with a test dipole
as the active element were repeated with real MTEs to check whether modern devices
are capable of adapting to a drastically changed environment.

If the passive element is excited by an MTE, the design of the antenna and the
distance determine the induced SAR. The measurements show, nevertheless, that all
tested devices remain below the case when the antenna touches the head (Figure
5.11).

When testing the devices according to the proposed WGMTE positions of CEN-
ELEC, the antenna never touches the head. By not taking the passive element into
account it is then theoretically possible that the actual absorption would be underes-
Figure 5.7: Determination of the resonant length of a thin rod which is attached to the head and irradiated by a dipole in 15 mm free space distance. The "worst case" - when the active dipole touches the head = 100%

Figure 5.8: Determination of the resonant length of a thin rod which is attached to the head and irradiated by a dipole in 15 mm free space distance. The "worst case - when no "passive" structure is present = 100%

estimated. Testing several MTEs showed that the measured SAR averaged over 1 cm³ without the resonant structure could lead to about 45% smaller absorption values than with the structure (Figure 5.12).
Figure 5.9: SAR value in the head averaged over 1cm³, when a resonant passive structure (100 mm) is attached to the head, depending on the distance of the parallely polarized active structure. The "worst case" - when the active dipole touches the head = 100%.

Figure 5.10: Real part of the feedpoint impedance of the active test dipole dependent on the distance from the body with and without the passive element.

However, induced SAR values which exceed safety limits due to an adjacent resonant structure would be highly unlikely. Only with parallel polarization and matching
Figure 5.11: Induced SAR values averaged over $1cm^3$ for the case of an adjacent resonant passive structure which is excited by four MTEs. The antenna and the passive structure have a parallel orientation. The values are normalized on the case where the antenna touches the head and are displayed as a function of the distance between head and antenna.

Figure 5.12: Induced SAR values averaged over $1cm^3$ for the case of an adjacent resonant passive structure which is excited by three MTEs. The antenna and the passive structure have a parallel orientation. The values are normalized based on the CENELEC positions with maximum absorption and are displayed dependent on the distance between head and antenna.
length can increased SAR values be observed. A twist from parallel polarization to 20° already leads to a 50% decrease in the SAR increase (Figure 5.13).

5.4.3 Internal Metallic Structures

Internal metallic elements such as head and dental implants excited by an external source could also cause increased SAR values. To study these effects, a simple setup was again chosen. This time a numerical tool was used to assess the SAR in tissue. Numerical investigations have the advantage that the tissue dependent resonant structures can be more easily determined and that possible excessive peak values can be better detected than with experimental means. The numerical tool was 3D MMP [9] [10]. The setup was the following:

A thin rod was excited by a plane wave at 900 MHz and 1.8 GHz in brain, bone and muscle tissue. For each tissue a resonant length of the thin rod was determined. The resulting induced SAR values were averaged over 1 g and 10 g tissue mass. In addition, the peak SAR value was determined. However, these values must be taken with caution, since the peak value is, nonetheless, "averaged" due to the computational discretization and it strongly depends upon the geometry of the implanted structure.

The results show that, at 900 MHz and averaged over 1 g, a fourfold relative increase in the SAR value for brain tissue and muscle tissue and a tenfold increase for bone tissue can be expected (Figure 5.14 and 5.15). This is due to the similar damping in brain and muscle (σ ~ 0.9 S/m) and a much lower damping in bone tissue (σ ~ 0.3 S/m). Since the increased SAR values are highly localized, the 10 g averaged SAR values are respectively lower (Figure 5.15). A tenfold relative increase in bone tissue must be seen in connection with bone tissue's much lower absorption. At 1.8 GHz the relative increase is only marginal, due to the stronger damping of the
Figure 5.14: Relative enhancement of the 1 g averaged SAR caused by an "implanted" metallic rod at 900 MHz as a function of the length of the thin rod (simulated for three tissue types: brain, bone and muscle).

Figure 5.15: Relative enhancement of the 10 g averaged SAR caused by an "implanted" metallic rod at 900 MHz as a function of the length of the thin rod (simulated for three tissue types: brain, bone and muscle).

A different situation is encountered when looking at the peak SAR values. For the case of (e.g.) a thin rod in bone tissue at 900 MHz, the relative increase reaches 500 times the of value when no implant is present! (Figure 5.16). The number must be
Figure 5.16: Relative enhancement of the peak SAR values caused by an "implanted" metallic rod at 900 MHz as a function of the length of the thin rod (simulated for three tissue types: brain, bone and muscle).

seen as rather arbitrary, because it depends on the geometry and the discretization. The magnitude of the enhancement, however, should be taken into account regarding the operational safety of mobile communication devices. The peak SAR induced by internal metallic structures can reach excessive peak levels.

The question of dental implants is of no serious concern. First, dental implants are usually too far from the source and second, the tissue in the mouth and cheek region is highly temperature resistant.

Regarding head implants, this worst case approach demonstrated that implanted metallic structures can lead to enhanced spatial SAR values and in particular to excessive peak values in the head. These effects are highly dependent on the geometry of the structure. Therefore, with respect to compliance tests it would be necessary that the actual exposure of the user with such an implant must be determined for every possible metallic structure. Assuming the numerical tools were available the expenditure would nevertheless be excessive: X-ray based determination of the shape of the implant, numerical modeling and worst case computation with all its restrictions. For compliance tests this is not feasible. It is therefore recommended that special considerations should be taken for persons with metallic head implants.

5.5 Conclusion

These results confirm the previous assumption that homogeneous models are superior to heterogeneous representations of the human head for the compliance testing of MTEs. Though homogeneous models generally overestimate the absorption in the
ear region, small shifts of the device in this region would cause large variations in the SAR when heterogeneously modeling the head. Heterogeneous modeling of this region would hence require several tests to guarantee that the maximum SAR was detected. A homogeneous simulation, on the other hand, guarantees that the highest possible value can be found with one single measurement. In addition, no enhanced SAR values are observed with an adjacent resonant structure which cannot be covered by homogeneous simulation. Whereas for heterogeneous modeling of such complex regions as around the ear, enhanced values must be expected. The latter also supports the homogeneous approach, since the hand holding the device can considerably change the current distribution on the device and the antenna and, therefore, the location of the SAR maximum in the head. Only implants cannot be covered and must be excluded from coverage by compliance testing.
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Chapter 6

Uncertainties and Critical Design Factors

Abstract - Based on the results of the previous chapters, an overall error and uncertainty estimation for a homogeneous model regarding the maximum volume averaged SAR can be performed. This includes uncertainties and errors related to the dosimetric measurement system, the simplification of the human body, uncertainties caused by metallic structures adjacent to the head and inside the head, the influence of the hand holding the device and the sensitivity of the positioning accuracy of the device with respect to the head. This uncertainty estimation enables the determination of the most critical design factors for a standard phantom model.

6.1 Estimation of Uncertainty

Homogeneous models as described in Chapters 3 and 4 usually overestimate the actual averaged SAR induced in the user. For compliance tests the uncertainties should be kept as small as possible, since all errors and uncertainties must be considered as an additional safety factor. Because preliminary tests [1], [2] showed that modern mobile telephones cause absorptions in the user which come close to today's safety limits, a quantification of these variations and, if feasible, a reduction of the uncertainties is necessary, if acting in the interests of both consumer safety and industry.

Based on the results from the previous chapters, the uncertainties can be quantified and the most sensitive factors regarding the maximum averaged SAR determined, as required for standardized compliance tests [3]. An overall uncertainty estimation for a "worst case" homogeneous model of the human head for compliance tests can be summarized and completed as follows:

- Uncertainties due to the assessment method. In the case of the chosen experimental approach, this includes calibration errors, errors in the probe's isotropy, errors due to a limited spatial resolution, etc. (see [4], [5] and Chapter 2).

- Error and uncertainties due to the simplification of the human anatomy. Influence of size and shape of the head. Role of the complex inner anatomy of the
head on the accuracy of the volume averaged SAR. Choice of the homogeneous simulating media (see Chapters 3 and 4).

- Error and uncertainties due to external and internal metallic structures, as well as the hand which holds the device (Chapter 5).

- Uncertainties and assessment errors which are directly tied to the device, such as manufacturing and specification tolerances. Finally, the sensitivity of the SAR on inaccurate relative positioning of the device in respect to the head [6].

6.1.1 SAR Assessment Method

In [4] and [5] the uncertainty of the assessment system DASY2 in reproducing a measurement is estimated to be smaller than ±5%. In Chapter 2 a new improved calibration method for E-field probes for use in dosimetric tests is described in detail and the total uncertainty was found to be within ±10%. Since the experimental measurement system determines the induced SAR in tissue simulating liquids, the assessment of the liquid's electrical parameters comes as an additional source of uncertainty. With the chosen methods the electrical parameters (i.e., $\sigma$) can be determined with an accuracy of ±6%.

The accuracy of the assessment of the electrical parameters of actual living tissues is difficult to determine, since data is either provided without information about error or uncertainty [7], [8], [9] or very low values are given, such as for example ±4% in the case of the conductivity [10], [11], [12]. Nevertheless, when comparing the electrical parameters for the same tissue type in existing literature, the conductivity can vary by ±25%. It is very difficult to characterize dielectric parameters in living humans. Usually, freshly excised animal tissues are used, which are subject to variations, due to condition or consistency of the material. Thus, samples of the same tissue type and under the same condition are taken and an averaged value determined. For compliance tests, which values are to be used for the simulating liquid should be agreed upon. For the uncertainty estimation this question must thus remain unsolved.

6.1.2 Anatomical Simplification

In Chapters 3 and 4 the influence of the SAR upon complex anatomical structures was investigated. Comparing the homogenized models with the heterogeneous complex model which shows the highest volume averaged SAR (i.e., "worst" complex model) yields an estimate of the homogenization error. Since "only" 3 complex head models were available it was not possible to determine a statistically "worst case" anatomy. However, it can be expected that the actual error would even be smaller (Figure 6.1).

The four numerical models used in Chapters 3 and 4 vary considerably in size and shape. Thus, if all are simulated homogeneously and with the identical exposure situation these homogeneous models enable an estimation of the influence of size and shape on the spatial SAR. The 4 numerical homogeneous models M1h, M2h, M3h and M4h show, at 900 MHz and for 1 cm$^3$ and 10 cm$^3$ averaging volumes, a deviation in the spatial SAR of less than ±10% with respect to the median value (Figures 3.14, 3.15), which is well within the numerical analysis uncertainty. As expected from the absorption mechanism the deviation is even smaller for higher frequencies (Figures...
Figure 6.1: Explanation of the error and uncertainties involved in the assessment of the SAR in various models. The "overestimation" is based on the worst complex numerical model. A smaller "overestimation" would be expected if a statistical analysis on the anatomical variations could be performed.

4.5, 4.6). Figure 6.1 describes the errors and uncertainties involved in the assessment of the SAR in appropriate head models.

6.1.3 External/Internal Structures

Increased SAR, due to external metallic structures adjacent to the human head, can lead to higher absorption values only under highly unlikely situations. Therefore, it can be assumed that the uncertainty is negligible in the case of mobile telephones. However, this question remains a problem for dispatch radios, which are operated in front of the head and at greater distances from the head.

Regarding internal metallic structures the results show that enhanced SAR values, in particular peak values, must be expected. The relative increase in SAR is strongly dependent on the geometry of the implant and, particularly in the case of the peak SAR, could reach nearly "any" value. Since such strong local field strengths inside the tissue are not considered in the safety standards, it is recommended that metallic implants should be excluded from this issue.

6.1.4 Device and Position

Three error sources are addressed with respect to the device and its position: first, variations due to the manufacturing process; second, variations of the specified output power; and third, the accuracy of the positioning of a device at the model.

The first and second error sources can easily be covered by testing samples from a given set of devices of the same type.
The positioning accuracy turns out to be most crucial. As already described in [6], the induced SAR drops with the square of the distance of the source. Depending on the design of the device (thickness, antenna mounting site, antenna type), the usual distance between radiating structure and head lies between 10 mm and 50 mm. Assuming 10 mm as a lower limit, a positioning inaccuracy of 1 mm would lead to an error of ±20%. Thus, the uncertainty is determined to be smaller than ±20% per millimeter positioning inaccuracy.

Additional factors, such as external metallic structures, are either negligible or must be excluded, as is the case with implants. Not modeling the hand which holds the device introduces an uncertainty of less than ±5%, which lies within the reproducibility of the dosimetric assessment system used for these investigations. Uncertainties regarding the device design and construction can easily be nullified by testing samples of one device type.

### 6.2 Critical Design Factors For a Standard Phantom

Based on this uncertainty estimation, it can be concluded that for compliance tests a shell model which is filled with tissue-simulating liquids is suitable to describe the maximum electromagnetic exposure of a user under the following conditions:

- The tissue simulating liquid should simulate the predominantly affected tissue types in the frequency bands of mobile communications between 100 MHz and 2.5 GHz. Numerical studies have been a good base to determine which tissues are predominantly affected depending on the frequency band and which should be simulated. The simulating liquid at room temperature must meet the electrical parameters of live tissue at body temperature.

- The shell must be as thin as possible (~2 mm) and should be made of material with low permittivities (\(\varepsilon_r < 5\)). The outer head geometry can be widely simplified, except for the dominantly exposed region above and behind the ear and the entire neck region. In these head regions the shape of the model should resemble as closely as possible the “worst case” real situation, i.e., the distances

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Table 6.1: Summary of uncertainties due to factors other than homogenization for the assessment of the maximum SAR. The positioning accuracy is crucial.
between the device, in particular between the antenna, and the head should correspond with the shortest distances that are encountered in reality.

- An accurate MTE mounting device should guarantee an exact placement of the device with respect to the shell.

Under these conditions, together with the dosimetric assessment system DASY 2 [4], a homogeneous model of the human head enables the assessment of the actual maximum averaged local SAR in the head of a user of an MTE with small errors and uncertainties.

6.3 Conclusion

The SAR measured in homogeneous phantom models can cover the maximum induced local averaged SAR in the head of a user with respect to all foreseeable uncertainties. When using an appropriately shaped shell model with an appropriate liquid, it is guaranteed that the device will encounter a situation similar to reality, regarding feedpoint impedance and current distribution. Homogeneous models are advantageous to heterogeneous, due to the fact that they cover uncertainties which come from external metallic structures, from small shifts of the device along the head surface and the hand holding the device. The over-estimation of a real "worst case" assessed in a homogeneous model is small. However, internal metallic structures, such as implants, must be excluded from coverage by a standard phantom model, due to the strong uncertainties which come with the unpredictable geometry effects of such structures.

A standard phantom based on these conclusions together with the dosimetric scanner DASY 2 [4], [5] fulfills the requirements of the CENELEC prENV 501066-2 [3] for standardized tests of mobile communication devices.
References


Epilogue

The availability of a dosimetric scanner (DASY), at a time when public concern about possible health hazards due to handheld mobile phones was great, helped to spread the idea of an experimental approach based on a homogeneous simulation of the human body to assess the absorbed power in the user. All major telecommunications companies purchased the scanner. Thus, DASY became a quasi standard within a short period of time. That not simply availability at the right time was the reason for its success but also the technological and scientific state-of-the-art of the entire setup was demonstrated when its initial purpose for safety testing was widened to include the design phase of a mobile phone: some 70% to 80% of the irradiated power can be absorbed in the head of the user of a mobile phone instead of being more practically transmitted to the next base station (Chapter 3). The crux of today's mobility, however, is and will remain battery time together with standby time in the case of mobile telephones. A reduction of the power dump into the head of the user and enhanced transmission for communication has consequently become an essential requirement in the design of efficient modern devices.

First devices with improved designs are currently entering the market. The electromagnetic exposure of persons using such devices is therefore reduced as well. Both factors have become a marketing issue in a highly competitive market, where profit margins are constantly under pressure. Consequently the more attractive device due to increased standby time and talking time is also the safer device which deposits less power into the head of the user.

However, the safety issue will remain a sensitive topic. This present study refers to today's safety limits, which correlate the hazard to thermal effects. The results of this study together with the dosimetric scanner DASY provide the scientific bases for compliance tests of mobile communication devices. A standard model incorporating the critical design factors as described in Chapter 6 is currently being manufactured. Thus, devices can be reliably and accurately tested as they come off the production line or already during their design process. How long it will take for standards organizations to agree to a standardized method for compliance tests is still open. The normative power of the facts will definitely help to accelerate the process. Nevertheless, the general risk to the public must be seen as minimal in comparison with other imminent hazards of modern society. However, the economic risks facing a company which does not take this issue seriously are very real, given a risk aware public and the strong competition. Those companies who reacted early to the problem of nearfield absorption were able to put improved devices on the market with better communication performance by minimizing induced SAR values.

Still unsolved is the problem of so-called athermal effects due to electromagnetic fields in the human body. These effects are assumed to occur below the absorbed power levels covered by today's safety limits and are seen to be caused by direct
interaction of electromagnetic fields with biological systems on the cellular level. An interaction mechanism is not yet known. This direct interaction has also occasioned on as-yet unproven association with the occurrence of cancer. Clues are coming from experiments using high-frequency fields which are amplitude modulated with extremely low frequencies. The multiple access method used for the worldwide GSM systems (see Chapter 1) causes signals of this kind. A proven, even weak, link between electromagnetic fields and cancer would be disastrous to industry. Thorough investigations are therefore necessary to clarify this issue. However, the outcome of these studies is still unclear and the spread of wireless communications continues unrestricted. Early knowledge, regardless of the outcome, enables appropriate steps, to respond at the earliest stage possible, to be taken.

To find an appropriate approach towards research in this field is crucial for both industry and public health organizations. Appropriate research policy would locate the best groups and laboratories in their specific field of expertise, would pool the interdisciplinary knowledge and allow independent research to be performed. Motorola's research policy, by choosing the best independent laboratories worldwide and targeting the investigations on specific crucial topics, could be a role model for other companies and organizations. The still widespread waste of enormous financial resources mainly in extensive PR-measures, based on incomplete scientific data, has proven to be inappropriate to deal in the long term with risks of a high technology consumer world. It neither serves industry nor consumer interests. Since safety issues will grow in importance, increased demand for financial resources requires a long term investment in appropriate risk assessment, analysis and management.

I conclude this thesis by expressing my hope that appropriate approaches will be taken to clarify the open issues regarding the safety of mobile communications.
Curriculum Vitae

Education

Dipl. El-Ing. ETH (Certified Electrical Engineer), degree 1991.
Swiss Federal Institute of Technology. Zürich, Switzerland.
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Matura Humanistisches Neusprachliches Gymnasium, degree 1983.
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