MEDICAL PHYSICS *International*

10th Anniversary of the IOMP Journal Medical Physics International

CONGRATULATION ADDRESS TO MPI JOPURNAL FROM JOURNAL PHYSICS MEDICA

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FIVE TIPS TO IMPROVE YOUR TEACHING OF MEDICAL PHYSICS

The AAPM Educators Resource Guide

Introducing Medical Physics to Students for Career Opportunities

MAPS FOR DEVELOPING MEDICAL PHYSICS CONCEPT NETWORKS IN THE MIND

The Encyclopaedia of Medical Physics II Edition

IAEA TRAINING RESOURCES ON RADIATION PROTECTION IN DENTAL RADIOLOGY

THE ASEAN DIAGNOSTIC REFERENCE LEVELS IN MEDICAL IMAGING

REPORT ON THE VIRTUAL MEETING OF 19th SEACOMP, 14th ACOMP and 13th TMPS

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TOWARDS POTENTIAL HARM ASSESSMENT FROM THE INDIVIDUAL PATIENT RADIATION DOSES IN IMAGING

PROFESSOR EHSAN SAMEI AWARDED THE MARIE SKLODOWSKA-CURIE AWARD OF IOMP, 2022

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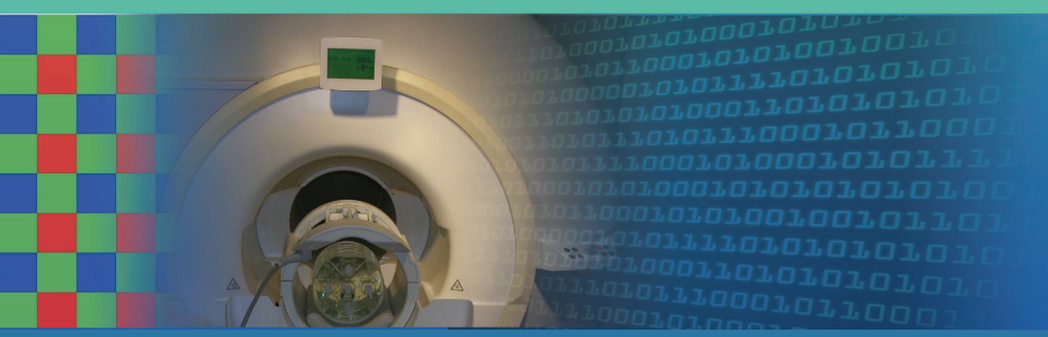
"INTRODUCTION TO MEDICAL PHYSICS"

PAUL LANGEVIN (1872-1946): THE FATHER OF ULTRASONICS

THE SCIENCE OF MEDICAL IMAGING - AN INTRODUCTION TO THE QUEST FOR VISIBILITY

Magnetic Resonance Imaging - Principles, Methods, and Techniques



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MPI

MEDICAL PHYSICS INTERNATIONAL

THE JOURNAL OF

THE INTERNATIONAL ORGANIZATION FOR MEDICAL PHYSICS

MEDICAL PHYSICS INTERNATIONAL Journal, vol.10, No.1, 2022

MEDICAL PHYSICS INTERNATIONAL
The Journal of the International Organization for Medical Physics

Aims and Coverage:
Medical Physics International (MPI) is the official IOMP journal. The journal provides a new platform for medical physicists to share their experience, ideas and new information generated from their work of scientific, educational and professional nature. The e- journal is available free of charge to IOMP members.

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EDITORIALS

10th Anniversary of the IOMP Journal *Medical Physics International*

Slavik Tabakov[1], Perry Sprawls[2]

[1] King's College London, UK, MPI Co-Editor in Chief
[2] Sprawls Educational Foundation, USA, MPI Co-Editor in Chief

Abstract: **The paper briefly describes the history of the Journal Medical Physics International (MPI) – the Journal of the International Organization for Medical Physics (IOMP), highlighting the milestones in its development and achievements in support of the global advancement of the medical physics profession.**

Keywords *– Medical Physics International Journal, Medical Physics Education, Medical Physics history.*

I. The beginning of the MPI journal

The history of the IOMP journal Medical Physics International (MPI) began in July 2012. The establishment of the MPI journal was initiated by Slavik Tabakov (at that time IOMP Vice-President) due to the necessity of a Journal to address e-learning in medical physics. This new type of education includes a number of applications with short life-cycles, hence a quick exchange of information about their application is essential. E-learning had already found a steady place in the profession, on one side - encouraged by the pioneering of e-learning in the profession and related EU Leonardo da Vince Award; on another side supported by the excellent IT skills in the profession.

The IOMP ExCom and the then President KY Cheung strongly supported the establishment of this Journal. The renowned medical physics educationalist Perry Sprawls was invited, together with Slavik Tabakov, as Co-Editors in Chief of the new Journal, with the IOMP ExCom members supporting them as members of the Editorial Board. At that stage it was decided that the journal would not only cover e-learning and education, but also other professional issues in medical physics. The sharing of expertise on these topics would support the global development of medical physics.
A distinguishing feature of the journal MPI was it does not publish peer-reviewed research reports as this is provided by the many other medical physics journals.
It was decided that the MPI journal will be free for all IOMP members and will be developed as an online e-Journal with normally 2 issues per year. The name of the Journal *Medical Physics International* was suggested by the then Chair of the IOMP Science Committee Willian Hendee.

During the summer of 2012 several activities started in parallel:
-The application and approval of the ISSN number (International Standard Serial Number) necessary for serial publications;
-The discussions with the other research-orientated medical physics journals, agreeing that MPI will complement their activities by covering predominantly topics related to education and professional issues and sharing of practical information.
-The development of a dedicated web site for the MPI Journal (an activity led by Magdalena Stoeva, then Associate Editor of the IOMP Newsletter Medical Physics World), plus securing a domain for the journal, agreed as: www.mpijournal.org

These activities were completed by the end of 2012 and the first issue of the Journal was prepared for the Spring of 2013 – in line with the celebrations of the Golden 50th Jubilee of IOMP during 2013 [1].

II. First issues and successes of the MPI journal

From its first issue MPI Journal became a very well accepted platform in the profession and although there were 2 issues per year, the monthly visitors of the web site were around 5000, sometimes up to 10,000 [2].

Gradually the Journal included more topics for sharing experience (some practical test objects or clinical procedures, which simplify the everyday tasks of medical physicists, etc.). Other topics were included, related to clinical practice – e.g. sharing experience with the industry of relevant medical equipment about its clinical use [3].

The papers related to e-learning and education had a significant number of readers and included lectures on specific topics which can be shared as free resources between the educators in the profession.
The medical physics profession already had a double growth in the decade 1995-2005 (4000 new medical physicists per decade, compared with the flat 2000 physicists per decade growth for 1965-1975, 1975-1985, 1985-1995). The decade 2005-2015 doubled the growth from 1995-2005, reaching 8000 per decade [4]. This trend continues and the current growth is expected to double again by 2025. Both the introduction of e-learning in the profession and the emphasis on education, supported by the MPI Journal, were significant drivers for this growth.

Fig.1 shows the online visits of the MPI web site at various periods in the past 10 years.

Fig.1 a. Number of MPI website visitors (per month) during the first 3 years of MPI (2013-2015) – *official website statistics*

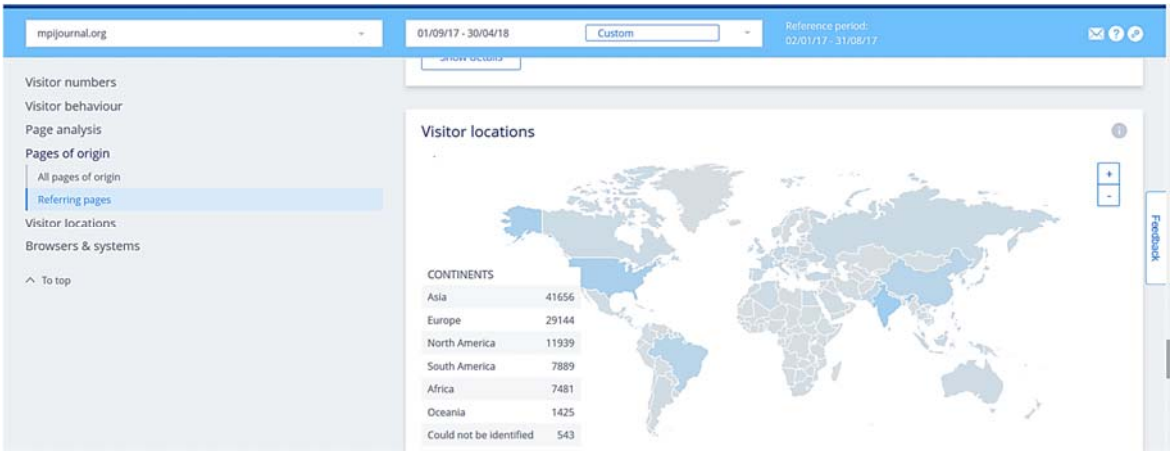

Fig.1.b. Distribution of MPI website visitors per continent for a period of 9 months: Sep 2017 to April 2018 (this period covers the time around one issue (starting after one issue has been published in June 2017, a second Issue has been published in Dec 2017 and before the third issue has been issued in May 2018) – *official website statistics*

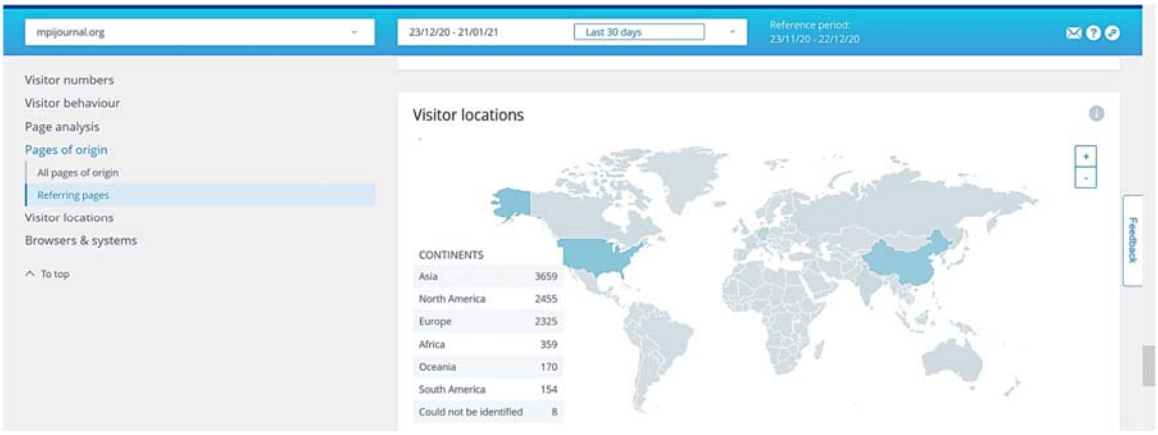

Fig.1.c. Distribution of MPI website visitors per continent for a period of 1 month: mid Dec 2020 to mid Jan 2021 (when a new MPI Journal was issued) - *official website statistics.*

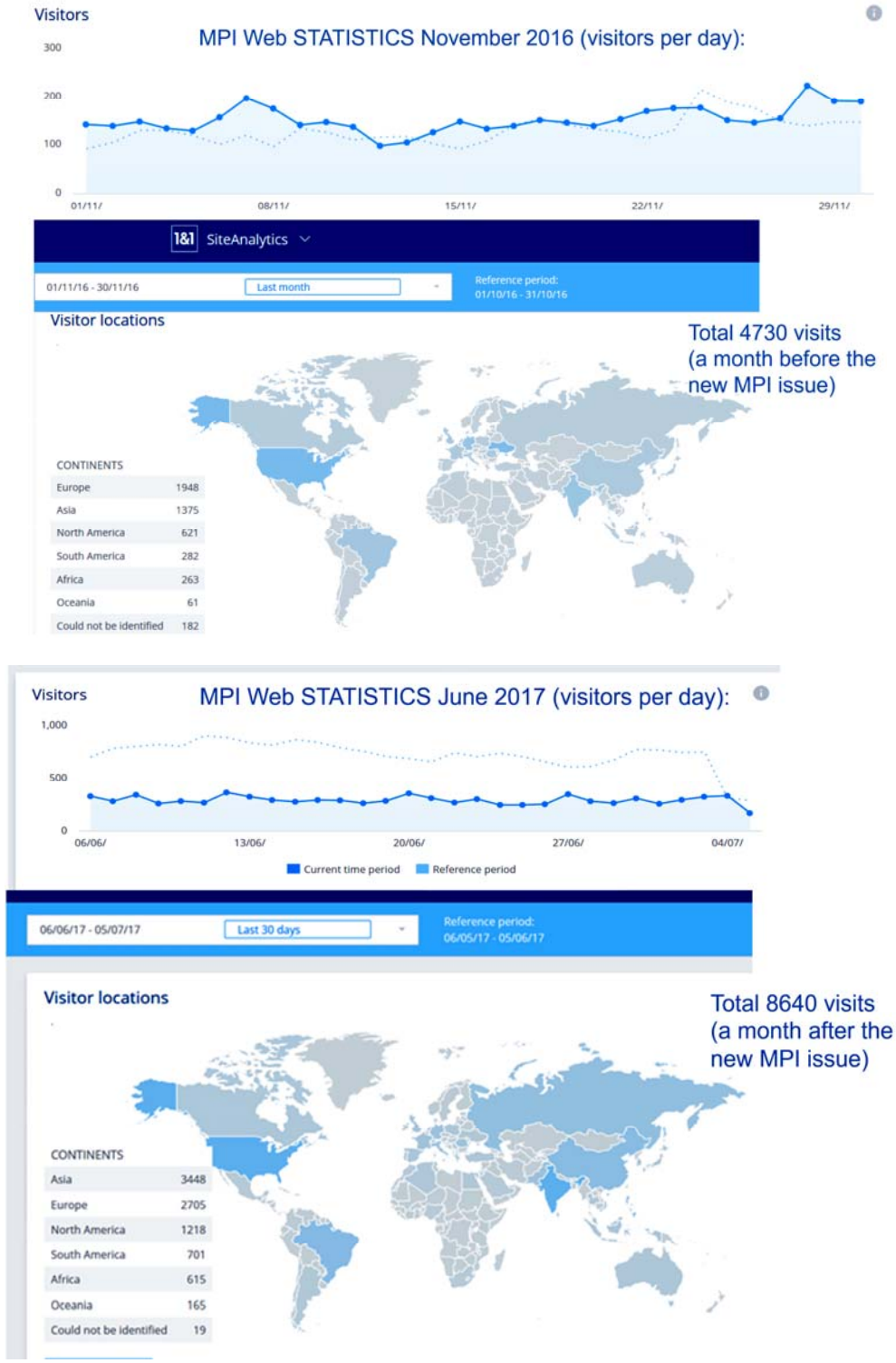

Fig.2. MPI visitors' statistics (worldwide) before and after publication of a new issue - *official website statistics.*

Fig.1 presents the official server statistic for the period between 2013 and 2021. Fig.1.a presents the visitors per month, while Fig.1.b, c - the geographical spread of MPI readers.

Fig.2 shows the MPI website visitors per day - one month before a new issue and one month immediately after a new issue has been published.

All papers in MPI are free to download. Some of these have thousands of downloads, and 6 papers have over 10,000 downloads.

III. MPI CONTENT DEVELOPMENT

Over time the content of MPI was enriched with various publications by colleagues from over 70 countries. Some abstracts of PhD theses were also published, thus monitoring the level of development in various countries.

From 2015 MPI began publishing specific papers about free educational resources, especially from the rich library of the AAPM (an activity led by P Sprawls). This was very useful for the colleagues from LMI countries.

From 2018 MPI started working with the IOMP Regional Organisations to collect information about the professional development from almost all National Members Organisations. This activity was led by S Tabakov, together with the Presidents and Secretaries General of all 6 RO. The volumes included information from 65 countries, distributed as follows:

in Latin America (ALFIM) from 2019 [5];
in Africa (FAMPO) from 2019 [6];
in South-East Asia (SEAFOMP) from 2020 [7];
in Asia and Oceania (AFOMP) from 2020 [8];
in the Middle East (MEFOMP) from 2021 [9];
in Europe (EFOMP) from 2021 [10].

From its first volume to the current volume 10, MPI has published 331 papers on education, professional developments, new resources and books, etc. These papers are about 1600 pages.

In addition to these papers MPI published various documents and guides from organizations (as IAEA), large international activities and MSc abstracts of the ICTP-University of Trieste international MSc programme with students from Low-and-Middle Income (LMI) countries [11]. These publications are about 400 pages.

The overall volume of the papers, published in MPI is about 2000 pages.

IV. MPI PUBLICATION OF CONFERENCE ABSTRACTS

From its 2nd issue MPI started to publish abstracts of the International Conferences on Medical Physics (ICMP). These were prepared by the Editors of the respective Conference. Together with these MPI published also the abstracts of the Conferences/Congresses of the IOMP Regional Organisations (RO) and related activities. So far the published abstracts are from:

-20th ICMP: Brighton, UK, September 1-4, 2013;
-RPM 2014: Varna, Bulgaria, 30 May-2 June 2014;
-22nd ICMP: Bangkok, Thailand, 9 – 12 December, 2016;
-24th ICMP: Santiago, Chile, 9 – 12 September, 2019
-AOCMP: Phuket, Thailand, 3-5 December 2020

By 2022 MPI has published about 2100 pages of abstracts from these international events.

V. MPI SPECIAL ISSUE ON HISTORY OF MEDICAL PHYSICS

In 2016 a project was initiated by S Tabakov aiming to cover the History of Medical Physics. This project was decided to be published as Special Issues, the first one being published in 2018. Initially Editors of the Series of publications on the subject were S Tabakov and P Sprawls and in 2020 G Ibbott joined the Editorial team. A special sub-website was developed for the History issues by the MPI Technical Editor M Stoeva (then Chair of the Medical Physics World Board) [12]. So far 7 Special Issues have been published covering:

Special issue 1 [13]: *X-ray Tubes Development; *Film-Screen Radiography Receptor Development; *History of Medical Physics e-Learning Introduction and First Steps

Special issue 2 [14]: *Fluoroscopic Technology from 1895 to 2019; *The Scientific and Technological Developments in Mammography; *Review of the Physics of Mammography

Special issue 3 [15]: *History of Dental Radiography ; *The History of Contrast Media Development in X-Ray Diagnostic Radiology; *Medical Physics Development in Africa

Special issue 4 [16]: *A Retrospective of Cobalt-60 Radiation Therapy; *The Many Steps and Evolution in the Development of Computed Tomography; *Medical Physics Development in South-East Asia; *History of Medical Physics Education and Training in Central and Eastern Europe

Special issue 5 [17]: * Ultrasound - the First 50 Years;*Measurement of Acoustic Pressure and Intensity Using Hydrophones;* Measurement of Acoustic Power and

Intensity Using Radiation Force; *Thermal Methods for Ultrasound Measurement development

Special Issue 6 [18] : *History of Medical Ultrasound - Imaging; *The Diasonograph Story; *Hewlett Packard - Innovations that Transformed Diagnostic Ultrasound Imaging; *History Of Doppler Ultrasound; *A History of HIFU Therapy

Special Issue 7 [19]: History of IOMP

These Special issues attract significant interest with thousands of readers. The overall volume of the MPI Special issues so far is over 900 pages.

VI. Conclusion

For the first 10 years of its existence the MPI journal proved to be a very useful resource for the medical physics profession. During this period of time the Journal published 20 regular issues and 7 Special issues. The overall volume of the publications is about 5000 pages. All of these issues supported the IOMP global activities for the development of medical physics by providing information and sharing experience among IOMP member societies.

The free information provided by MPI to all IOMP members is considered by all colleagues as a very important contribution to their professional development. This is especially true for the colleagues in LMI countries.

The journal provided many papers and shared experience important during the Covid 19 pandemic. At that time the MPI Editors led a survey among many of the medical physics journals to assess activities of the publications in support of the unexpected change in professional and everyday life. All Journals reported increased activities, what was very important for the continuation of the activities in the medical physics profession. During the period June-December 2020 MPI had c.48,000 visits to its web site (c.37% from N.America, 28% from Asia and 27% from Europe) [20].

The steady number of MPI readers and the high number of downloads shows that the Journal content is very useful to the global community of medical physicists and that the educational and professional topics are of special importance for the profession.

Acknowledgements

The MPI Editors-in-Chief are grateful to all authors and invited editors who contributed to the Journal, the MPI Editorial Board, the Editorial assistant V Tabakova and especially the Technical Editor M Stoeva.

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CONGRATULATION ADDRESS TO MPI JOPURNAL FROM JOURNAL PHYSICS MEDICA – EUROPEAN JOURNAL OF MEDICAL PHYSICS

Iuliana Toma-Dasu [1]

[1] Editor-in-Chief of Physics Medica – European Journal of Medical Physics

ON THE 10 YEARS ANNIVERSARY OF THE IOMP JOURNAL MEDICAL PHYSICS INTERNATIONAL (MPI)

It brings me great joy to congratulate the IOMP Journal Medical Physics International (MPI) on its 10 Years Anniversary! I am honoured to have the opportunity to acknowledge on behalf of the Editorial Board of Physica Medica - European Journal of Medical Physics and its readers, the contribution of MPI to the progress in the field of medical physics, to the development of the medical physicist profession and to the growth of our international community of medical physicists.

With strong connection to IOMP, MPI is publishing articles of global interest for medical physicists on educational and training aspects, professional issues, research and development complementing in the optimal way the other medical physics journals in the spirit of collaboration fostered by the highly dedicated editors of MPI, Prof. Slavik Tabakov and Prof. Perry Sprawls. Indeed, the editorial policy of MPI focusing on identifying topics of high relevance for all the medical physicists around the world has shaped the profile of the journal and strengthened its position among the medical physics journals. Furthermore, the series of papers dedicated to the history of medical physics presenting not only the evolution and the milestones of the technology used in medical physics applications but also the overall development of the profession brought together our community in a unique way that is highly appreciated by the readers of the other medical journals, such as Physica Medica.

I would therefore like to acknowledge one more time the privilege of collaborating with MPI to serve the medical physics community, to thank its editors for their fantastic work to develop the journal and to make it available to everyone, and to whish them many more occasions of celebration.

We will be there, next to you, as always, to share the work and the joy.

Congratulation!

Prof. Iuliana Toma-Dasu, Editor-in-Chief of Physics Medica – European Journal of Medical Physics

CONGRATULATION ADDRESS TO MPI JOPURNAL FROM JOURNAL HEALTH AND TECHNOLOGY - THE OFFICIAL JOURNAL OF IUPESM

Magdalena Stoeva & Kang Ping Lin[1]

[1] Editors-in-Chief of Health and Technology - The official journal of IUPESM

MPI co-editors Dr. Tabakov and Dr. Sprawls succeeded in soliciting an excellent editorial team supported by some of the world's leading medical physicists.

Health and Technology editorial board acknowledges MPI's contribution to the various aspects of IOMP and medical physics – professional, education and training, collaboration, technology, global outreach.

Congratulations to a decade dedicated to Medical Physics!

Prof. Magdalena Stoeva and Prof. Kang Ping Lin Editors-in-Chief of Journal Health and Technology

10 YEARS IOMP JOURNAL MEDICAL PHYSICS INTERNATIONAL (MPI)

It is our pleasure and honour to congratulate the IOMP Journal Medical Physics International (MPI) on its 10 Years Anniversary!

10 years period is the first punctuation of growth when it comes for human age, but a decade in the professional world is an important milestone.

MPI's contribution to the professional aspects of medical physics is indisputable. The broad regional and functional coverage, the always up-to-date topics and the attention paid to the detail turned MPI into a tribune of medical physics profession throughout the years. MPI reaches out to medical physics societies and to individual medical physicists throughput the world through carefully curated works on various topics related to the present, past and the future of medical physics. Special focus is set on the History of Medical Physics turning these special series into unique source of information on how our profession developed technologically and organizationally.

EDUCATIONAL TOPICS

FIVE TIPS TO IMPROVE YOUR TEACHING OF MEDICAL PHYSICS

George Starkschall, PhD, FAAPM, FACR

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Keywords *– Medical Physics, Education*

Teaching is a significant component of the responsibilities of many medical physicists, but few medical physicists have received any training in how to teach. While requiring medical physicists to receive a degree in education is not practical, some training on how to teach can be very useful. The purpose of this paper is to identify some ways in which medical physicists can improve their teaching.

Because this is a paper on the topic of education, I would like to start it with a brief quiz:

- **Question 1**: How are your present medical physics practice methodologies different from those you used five years ago?

Question 1 should be relatively easy to answer, given the many changes that have occurred in medical physics over the past five years, including new treatment and imaging modalities, as well as new types of equipment.
Here is Question 2. Question 2 might be a bit more difficult to answer.

- **Question 2**: How are your present teaching methodologies different from those you used five years ago?

In order to respond to Question 2, some medical physicists may update their lecture notes from year to year, but for many, the course content does not change, nor do the lecture notes. After all, not much has changed in the Compton Effect as it relates to medical physics since the derivation of the Klein-Nishina formula in 1928 [1]. I recall when I was a graduate student, one of my physics professors, a distinguished Nobel laureate, lectured from the same worn notebook he had used for at least twenty-five years, if not more. Needless to say, his lectures were not very stimulating.

And what happens during these lectures?

The professor writes their notes on the blackboard (or displays them as PowerPoint slides) and the students carefully copy the notes into their notebooks.

However, as one commentator remarked: "In the digital world, there is no longer any reason to use class time to transfer the notes of the instructor to the notes of the student (without passing through the brain of either)." [2].

In order to enable the medical physicist to answer the second question in a positive manner, I would like to suggest several ways in which the medical physicist teacher (or any other teacher) might modify their teaching methodologies to improve the quality of their teaching. So, here goes:

1. Do not be afraid to use novel teaching methodologies in your classroom.

Novel teaching methodologies such as flipped learning and problem-based learning move away from the traditional lecture format to more interactive means of teaching. In applying flipped learning, for example, one records the lecture in which information is provided to the student. The student listens to the lecture prior to coming to class, and the classroom time is used to work out problems, answer questions, and provide insight into the material presented in the lecture. In my own experience with flipped learning, I was able to ensure that the students listened to the lecture by giving them a short, online quiz on the lecture material prior to their classroom meeting.

In applying problem-based learning to the classroom, a problem is presented to the student, and, in order to solve the problem, the student has to learn the subject material. In that way, the student can immediately recognize the value of the information they are seeking and how that information is used to solve a problem.

2. Test students' understanding of concepts in addition to their ability to solve problems

All too often, a student learns new material in order to solve a problem set or demonstrate knowledge on an examination, but fails to understand the concept behind the information. Eric Mazur, a Harvard physics professor, makes the point that students come to class with a preconceived notion of how the world works, and often new material presented in class violates their preconceived notions [3]. Mazur continues that the best individual to aid the student overcome these perceptions is not the teacher, who has worked with the knowledge for many years, but the student

who has recently overcome these perceptions. Mazur thus advocates the use of fellow students to assist each other in understanding concepts presented in the classroom.

3. Continually monitor students' understanding of material presented.

One property of physics knowledge is that it is progressive, that is, understanding concept n+1 is based on understanding concept n. Conversely, if a physics student does not understand concept n, it is not likely they will understand concept n+1. Thus it is important for the physics teacher to verify that the student does not get lost in following the class. Continually asking students questions about the class material is one way for the instructor to verify that they are not leaving the class behind. Another method uses an audience response system (ARS) either to enable the student to answer questions posed by the instructor or to signal the instructor that the student has lost track of the subject material.

4. Teach students how to guess input when precise input is not available

How many physics problems given to students are of the "plug 'n' chug' variety? If the student knows (or can guess) the correct formula, all they have to do is plug the given input values into the formula, and out comes an answer. The real world is not generally that simple. The student should learn to seek input information if it is not given. If they are unable to locate the input information, they should be able to provide a good estimate as to the correct value, but be able to estimate the uncertainty of the resulting answer. The standard example of such a problem is the classic Fermi problem, "How many piano tuners are there in (name of city)?" It is quite surprising to find that many students enter a graduate medical physics program completely unable to move forward in solving such a problem.

5. Base grading on performance expectations and not on how other students perform

So-called "grading on the curve" is unfair to students, placing them in competition with one another. A course grade is a measure of the extent of course material the student has learned. Given a specific amount of course material learned, what justification is there to base a student's grade on the amount of course material learned by other students? Identify a set of learning objectives and evaluate the student based on the extent to which the student meets those objectives. If many of students do poorly, it means that either the expectations are too high or the instructor did not do a good job of teaching. Do not change the standards to meet the students' performance after the fact [4].

Incorporation of the tips that have been presented may take medical physics teachers out of their comfort zones, but the way we learn is by extending the boundaries of our comfort zones. In the clinic, we venture out of our comfort zone any time we implement a new technology; in an analogous manner we need to venture out of our comfort zone when we implement a new methodology in our teaching.

Most of all, we need to remember that our role as medical physics teachers is not to teach our students medical physics. Our role as medical physics teachers is to teach our students to learn medical physics.

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The AAPM Educators Resource Guide

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Keywords *– Textbooks, Lecture Materials, Continuing Education, Teaching Resources.*

I. Introduction

Medical physics education is a complex process consisting of many elements. Central to the process is the medical physics educator (teacher, mentor, program director) who uses his or her knowledge and experience in providing effective and efficient learning activities. These activities include class and conference presentations and discussions, small peer group projects, and a variety of independent self-study engagements by learners. Each of these activities can be enriched with appropriate resources that support and enhance human learning and teaching activities.

The Educators Resource Guide (ERG) is a directory identifying resources that can be used by medical physics educators in a variety of educational activities, including classes in Academic Institutions, Continuing Education, and Individual Self Study. The ERG is hosted by the AAPM and all medical physics educators are invited to contribute content to the ERG. Content in the ERG is reviewed and updated by the Educator's Resource Guide Working Group (ERGWG), a subcommittee of AAPM Education Council

II. Content

The ERG is a directory of materials that are published and available from other sources and does not contain the actual resources. While some resources, especially textbooks can be purchased from the publishers, the emphasis of the ERG is to identify and provide links to the extensive collection of open access and free resources.

III. Avalibility

The ERG is an open resource and available for all to use at: https://www.aapm.org/education/ERG/

IV. Using the ERG

Content in the ERG is categorized by topic into the following major sections:

1. General resources
2. Resources to help medical physicists become better educators
3. Resources organized by category of learner

Please email the ERGWG (2022.ERGWG@aapm.org) with suggestions for additional content or edits to existing pages.

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Introducing Medical Physics to Students for Career Opportunities

Perry Sprawls

Emory University School of Medicine, Atlanta and Sprawls Educational Foundation, www.sprawls.org .

Keywords – *Medical Physics, Career*

As students advance through their academic programs, even with their major field of study decided, physics for example, there are often open questions about specific topics and areas of specialization to consider. Within the broad field of physics there are many. There are a variety of reasons why each of us chose our areas of work, Perhaps it was an admired professor and tutor, research opportunities, or perhaps the popularity and excitement of a field, astrophysics for example.

Knowledge of specific fields of specialization can be of significant value to students in making choices that will provide directions for their careers.

Those of us who are medical physicists have the opportunity, and perhaps the professional obligation to help students learn about "what we do" and the satisfaction it brings.. In some communities and institutions medical physics is not as highly visible as some other fields, astronomy, climate physics, etc. We are off in the hospitals and clinics doing our work.

We each have our opportunities to help students learn, and perhaps get interested and excited about medical physics. This includes having students come in and observe our activities, giving presentations to student groups, and maybe encourage academic programs, especially physics departments, to have seminars on medical physics topics.

Another opportunity that is provided here is an online document, like a book chapter, that can be distributed and passed on to students for them to explore. It describes one specific area of medical physics, imaging, which is the author's specialization. Perhaps later than can be a similar article on radiation oncology.

For now, let's use this article in the Appendix along with our personal activities to help students learn about our interesting profession of medical physics.

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ATTACMENT

MAPS FOR DEVELOPING MEDICAL PHYSICS CONCEPT NETWORKS IN THE MIND

P. Sprawls

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Abstract— Medical Physics Knowledge is generally a mental representation of the physical universe (instruments, equipment, interactions, procedures, etc.) in a combination **of** mathematical and verbal symbolic representations and sensory concepts. Both are required for the effective application of physics in clinical medicine. It is the sensory concepts that are the foundation of medical physics knowledge. Conceptual knowledge is most effectively developed through sensory interactions with the physical universe, especially visualization directly or with images. Conceptual knowledge can be developed from textbooks and in classrooms with appropriate presentations. However, this is often as a series of individual concepts or relatively small concept clusters. Concept maps can be used at the conclusion of a learning activity (class or textbook chapter) to unify the concepts into a comprehensive concept network that represents a higher level of learning and understanding. The creation and publication of high-quality concept naps provides an opportunity for medical physicists to gain recognition and contribute to the enhancement of medical physics education around the world.

Keywords— **Concepts, Mind Maps, Concept Maps, Sequential Learning, Physics Knowledge.**

I. INTRODUCTION

Our knowledge of medical physics and related topics is composed of a complex network of sensory concepts, especially visual, in the mind along with symbolic representations including verbal descriptions (words) and quantitative relationships with mathematical symbols and equations. Both types of knowledge, *conceptual* and *symbolic*, are required for the practice of medical physics, especially clinical applications. It is the network of concepts that is the critical knowledge for many medical physics activities. Sensory concepts are representations of the physical universe within the human mind and are generally developed by observing and interacting with the physical universe and can support many functions as illustrated in Figure 1.

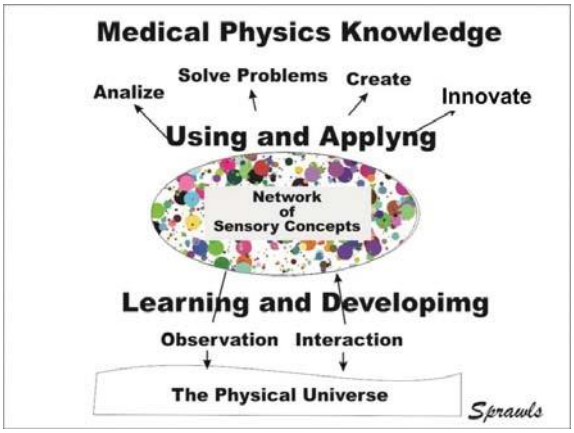

Figure 1. The significant role of sensory concepts in medical physics knowledge.

Let's begin by considering our knowledge of the physics of water. When and how did we develop it? Not in a classroom but by many interactions and observations as illustrated in Figure 2.

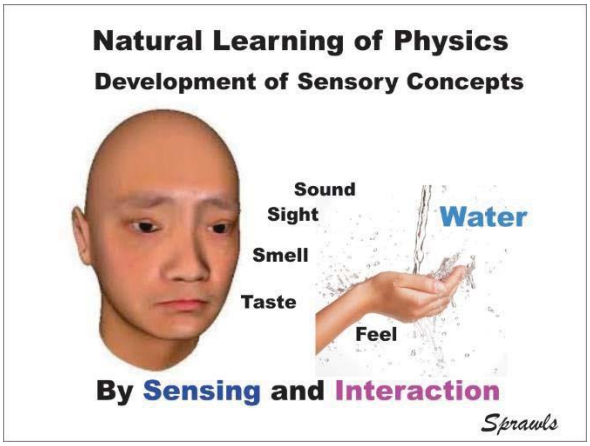

Figure 2. Learning the physical characteristics of water.

Perhaps in a physics class we learned some of the quantitative relations, hydrostatics, and hydrodynamics, but our most valuable and useful knowledge of water is the network of concepts developed through observations and interactions as illustrated. This is the knowledge that can support many activities throughout our lives.

Now let's consider knowledge of medical physics. As with our knowledge of water, it is the network of sensory concepts that supports many medical physics activities and functions as illustrated in Figure 3.

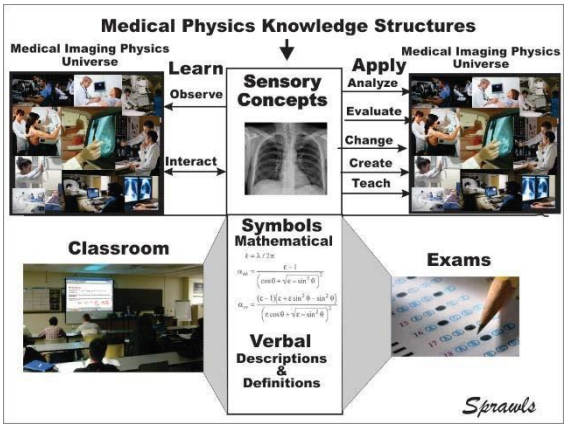

Figure 3. Two types of medical physics knowledge structures; sensory concepts and symbolic representations.

All representations are significant to the practice and applications of medical physics but in different ways. Symbolic representations, especially mathematical as generally taught in the classroom are useful for understanding quantitative relationships and making calculations. It is the type of knowledge most often evaluated with tests and examinations which gives it special emphasis in classroom teaching.

However, it is conceptual knowledge that is the foundation and unification that supports continuing learning and effective application of physics knowledge (Ref. 1).

This knowledge exists as a complex network of many individual concepts. In addition to understanding the individual concepts, the relationship among concepts, *the network of sensory concepts*, is a critical part of our medical physics knowledge.

As we observed with the physics of water, the development of a network of sensory concepts is a natural learning process occurring as we observe and interact with segments of the physical universe.

Concepts are the mental representations of the physical universe developed as we observe and interact with our senses, vision, feeling, hearing, etc. A concept generally includes our understanding of an item or subject and its characteristics.

There are two major characteristics of conceptual knowledge that distinguishes it from symbolic knowledge (words and mathematical symbols). It is a form of natural learning as we interact with and observe the physical universe, sometimes in the classroom or laboratory but often in other experiences. The great value of conceptual knowledge is it supports many activities when interacting with the physical universe, ranging from everyday things we do to advanced medical physics functions. Evaluating image characteristics and quality, performing radiation safety procedures, and conducting laboratory experiments are examples.

Our minds are filled with perhaps thousands of concepts of objects, events, and conditions that we have encountered. While an individual concept, *Water* for example, can be very useful knowledge, it is the *network of inter-related concepts* that is knowledge required for many applications.

Concept Maps (Ref. 2) as discussed here, are teaching and learning resources that when added to the more conventional teaching and learning methods can contribute to the formation of highly valuable concept networks in the learner's mind.

II. Sequential Teaching and Learning

Teaching and learning are generally sequential activities in which a series of individual topics are presented and learned as illustrated in Figure 4. This includes classroom presentations, textbooks, and modules.

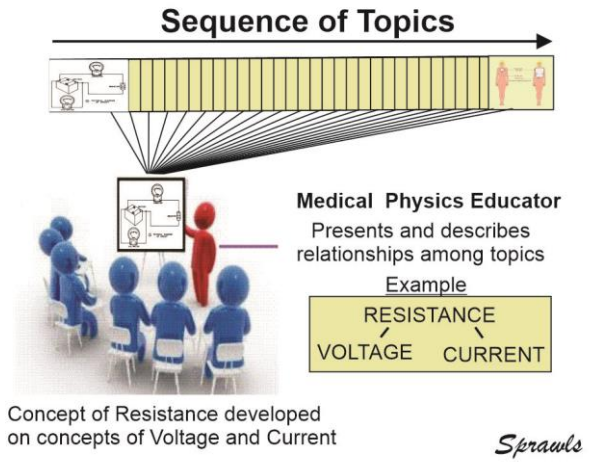

Figure 4. Sequential learning and teaching.

With sequential teaching and learning (Ref.3) and in textbooks, topics are often connected in a continuing series building on each other. As concepts are developed by the learners they are linked to other concepts. An example: to develop an appropriate concept of electrical resistance requires established concepts of voltage and current. This is to develop a useful concept of resistance, very different from memorizing Ohm's Law.

Most concepts developed in sequential learning activities are connected to other concepts; resistance is an example. These are generally small clusters of concepts as illustrated in Figure 5.

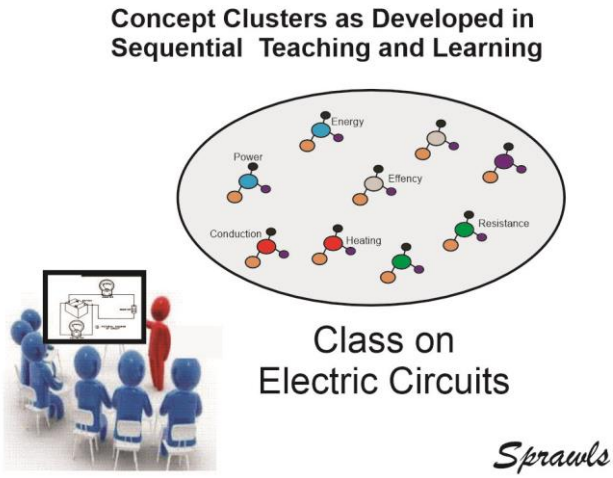

Figure 5. The development of concept clusters in classroom activities.

As learners develop concepts of a specific topic, for example resistance, it is generally linked to other concepts, including voltage and current. This can be considered as a *cluster* of concepts. It is a small network but not yet included in the larger concept network of medical physics knowledge.

In typical sequential learning activities, classrooms and textbooks, concepts are developed in clusters because of the way the information is presented and the relatively short attention and memory span of the human mind. As a learning activity, class presentation or reading a textbook progress, attention is focused on additional topics in sequence. Concepts and concept clusters are formed but not integrated into the comprehensive concept network.

That is what can be achieved with *concept maps* to unify the many individual concepts and provide the "big picture" of medical physics.

III. CONCEPT NETWORKS AND MAPS

A comprehensive network of concepts is a significantly *higher level of medical physics knowledge* compared to many individual concepts and concept clusters. It supports many medical physics activities, especially relating to clinical procedures and teaching.

The development of an effective concept network is a specific activity summarizing and integrating concepts learned during a class presentation or textbook chapter. A concept map as illustrated in Figure 6 is an example.

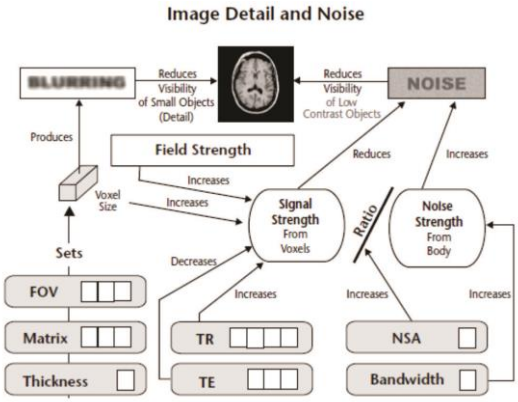

Figure 6. Concept map related to MR image detail and noise.

A major medical physics activity is evaluating and optimizing image quality for the various imaging modalities and methods. MRI is an example. It is generally a complex process because of the many variable conditions and factors involved. The concept map in Figure 6 identifies the factors and relationships that affect the two image quality characteristics, detail and noise, in MRI. The reality is all of these must be considered to fully understand and optimize image quality.

Concept maps can be used at the end or conclusion of a class or textbook chapter to connect the concepts into a network to enhance learning and for future reference.

The concept map in Figure 6 is from the chapter on Image Detail and Noise in the Sprawls textbook provided in the Addendum.

Concept Maps are a category of Mind Maps that are used extensively in education and many other activities.

IV. MIND MAPS

Mind Maps (Ref.4) are diagrams used to show relationships among topics, objects, events, conditions, ideas, etc. They have many applications, education being one of them. Tony Buzan, a British innovator and author, is generally recognized as the creator of mind maps and the promotor over the years. Figure 7 is a mind map created to illustrate his many activities.

Figure 7. A mind map showing the activities and achievements of Tony Buzan.

This is used here to illustrate the features and characteristics of mind maps. They are usually hierarchical in nature, branching out from one specific item or topic. In this example it is Tony. A major value of mind maps is they present the "big picture" associated with a topic and its many relationships.

Mind maps, or more specifically concept maps, have been used over the years by the author as a concluding summary of textbook chapters, classroom discussions, etc. More on that later. For a general review of mind maps see:

https://en.wikipedia.org/wiki/Mind_map

V. DEVELOPING AND USING CONCEPT MAPS

A concept map is an illustration or diagram of a specific topic showing relationships to other topics represented by mental concepts. The assumption is that learners have developed concepts or concept clusters of the individual topics in a class, reading and studying, or direct interactions with equipment, instruments, and procedures. The concept map is then used to connect and understand the relationships of the concepts in the mind.

Concept maps can be developed with graphics software used for producing classroom or conference visuals (author's preference) or with software specific for concept maps available on the internet by searching for "concept maps templates".

For medical physics educators, developing concept naps is a creative (and perhaps learning) process requiring thought about the organization and relationships of the topics. Developing good concept maps and publishing them for others to use will give recognition to the creators and will enhance the teaching of those who use them.

An example of published concept/mind maps are those by the author in the Addendum.

VI. SUMMARY AND CONCLUSION

Figure 8, Provides an overview and summary to conclude our consideration of concept networks and concept maps.

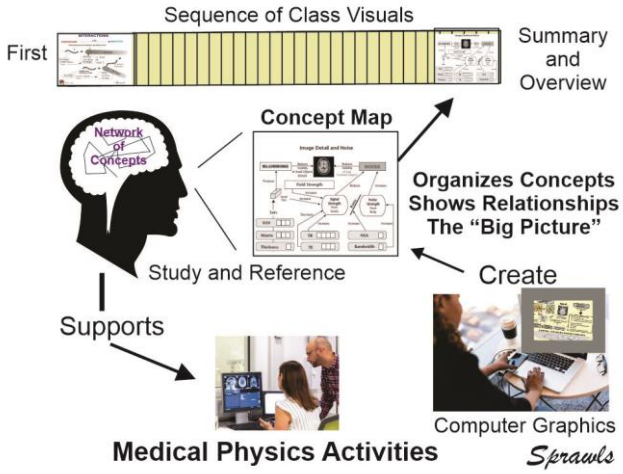

Figure 8. A Concept Map relating concepts associated with Concept Maps.

A comprehensive network of connected sensory concepts provides the medical physics knowledge that supports many medical physics activities. This is especially significant for interacting directly with medical equipment and procedures. Individual concepts and concept clusters are developed in classrooms, laboratories, studying textbooks, and continuing experience. However, these activities do not effectively develop comprehensive concept networks that are a higher level of learning. Including concept maps in educational presentations and discussions, modules, textbooks, etc. enhances the teaching and learning process to achieve this higher level of learning.

When used as a conclusion of a learning activity concept maps organize individual concepts and concept clusters into a comprehensive network with connections and relationships.

Using computer graphics to create and publish high-quality concept maps for others to use provides an opportunity for medical physics educators to use their knowledge and experience to enhance medical physics education around the world.

VII. REFERENCES TO EXPLORE CONCEPTS

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IX. ACKNOWLEDGEMENT

Dr. Debra Monticciolo, MD, Department of Radiology, Baylor Scott & White Healthcare - Central Texas provides the clinical radiologist perspective and collaboration in our continuing development of clinically focused physics education.

About the Author: Perry Sprawls is a clinical medical physicist specializing in diagnostic radiology and medical physics education. He is Distinguished Emeritus Professor at Emory University School of Medicine in Atlanta and now contributes to medical physics education around the world through the Sprawls Educational Foundation, www.sprawls.org. It is the combination of his experience as a clinical physicist and educator that is the foundation for developing and sharing resources to support the teaching of medical physics. His continuing research and development activities are resulting in models for increasing the effectiveness of both the learning and teaching process, especially for clinically applied medical physics. Throughout his career he has used mind and concept maps, both in classes and textbooks. The current effort described in this article is to illustrate the value and encourage the creation and use of concept maps to enhance medical physics learning.

Dr. Perry Sprawls

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ADDENDUM

Sprawls Magnetic Resonance Imaging P

The Encyclopaedia of Medical Physics II Edition: The update of General Terms field

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Abstract: **The paper describes briefly the update of the Medical Physics Thesaurus of terms and the related update of General Terms field of the Encyclopaedia of Medical Physics II Edition (published 2021).**

Keywords *– Medical Physics Encyclopaedia, Medical Physics Education, Medical Physics resources.*

I. Introduction

The Encyclopaedia of Medical Physics development and its update took over 20 years. The articles of the Encyclopaedia I Edition (published by CRC Press in 2013 [1]) were based on the Medical Physics Thesaurus of terms, developed in 2003 and updated in 2008. An additional minor update of the full Thesaurus was made in 2011. The Encyclopaedia Edition I included over 2800 articles explaining the foundation terms in medical physics. Some of those terms were General terms, supporting the medical physics knowledge. Edition I was published by CRC Press (in paper) as a two-volume set and uploaded (together with the Scientific Dictionary of Medical Physics Terms in 32 languages) on the dedicated website www.emitel2.eu as a free reference and educational resource.

During the following 10 years materials for the Thesaurus update were collected and a major update was made in the period 2019-2020. This update included about 650 new terms plus additional diagrams, tables and other information. The Encyclopaedia II Edition is listed alphabetically, but it has specific parts (fields), managed by different teams, as per the narrow specialty of the contributors. These fields are on Physics of: Diagnostic Radiology; Radiotherapy; Nuclear Medicine; Ultrasound Imaging; Magnetic Resonance Imaging; Radiation Protection; Non-ionising radiation protection; General terms (including Management). This new II Edition of the Encyclopaedia of Medical Physics was printed and published by CRC Press in 2021 [2]. The materials from the update were uploaded at the same website: www.emitel2.eu

II. General terms update

This paper describes briefly the nature of the update and the new General terms of the Encyclopaedia. The initial Thesaurus of Diagnostic Radiology terms (from 2003) included General terms related to mathematics, physics, medicine, materials, engineering, etc., supporting the knowledge database of medical physics.

The mathematical terms, related to medical physics included topics covering image reconstruction and processing, statistics, etc. All these are covered in specific publications, but their existence in the Encyclopaedia intended to give condensed knowledge and starting point for further studying of the subject (this was additionally supported by lists with Further Reading at the end of each encyclopedic article). Many of these terms were updated in Edition II. Care was taken the explanation of these General terms to be with educational value. Fig. 1 shows part of one such article.

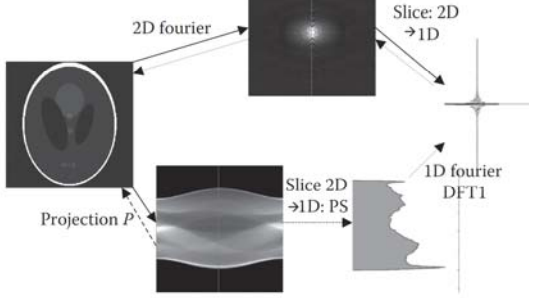
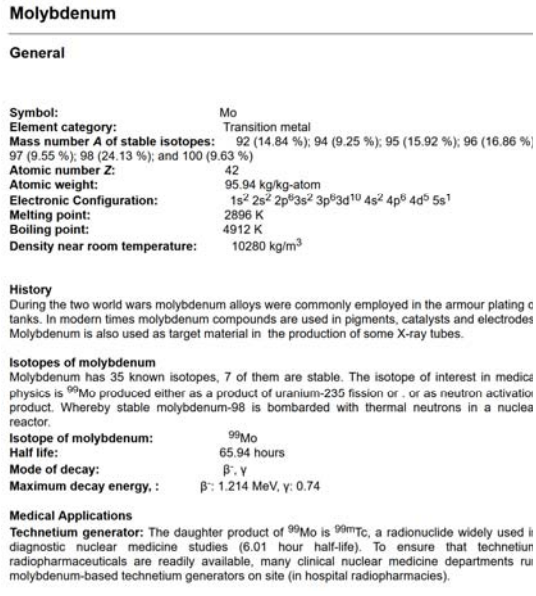


Fig.1 Graphical example for image reconstrction – *from article: Back projection reconstruction*

Most of the General terms from medicine were covered in Edition I (such as radiographic projections, parameters of some organs – e.g. densitometric values, etc.), but still some additional terms and values were added in Edition II.

The terms related to physics included many articles covering specific elements used in medical physics. These terms were updated, where necessary. Figure 2 shows a typical description of one element (Molybdenum) with the included parameters. The Encyclopaedia included also an addendum with fundamental physics constants and the Periodic Table of the Elements.



Molybdenum

General

Symbol:	Mo
Element category:	Transition metal
Mass number *A* of stable isotopes:	92 (14.84 %); 94 (9.25 %); 95 (15.92 %); 96 (16.86 %) 97 (9.55 %); 98 (24.13 %); and 100 (9.63 %)
Atomic number *Z*:	42
Atomic weight:	95.94 kg/kg-atom
Electronic Configuration:	$1s^2\ 2s^2\ 2p^6 3s^2\ 3p^6 3d^{10}\ 4s^2\ 4p^6\ 4d^5\ 5s^1$
Melting point:	2896 K
Boiling point:	4912 K
Density near room temperature:	10280 kg/m^3

History

During the two world wars molybdenum alloys were commonly employed in the armour plating c tanks. In modern times molybdenum compounds are used in pigments, catalysts and electrodes Molybdenum is also used as target material in the production of some X-ray tubes.

Isotopes of molybdenum

Molybdenum has 35 known isotopes, 7 of them are stable. The isotope of interest in medica physics is ^{99}Mo produced either as a product of uranium-235 fission or . or as neutron activatio product. Whereby stable molybdenum-98 is bombarded with thermal neutrons in a nuclea reactor.

Isotope of molybdenum:	^{99}Mo
Half life:	65.94 hours
Mode of decay:	β^-, γ
Maximum decay energy, :	β^-: 1.214 MeV, γ: 0.74

Medical Applications

Technetium generator: The daughter product of ^{99}Mo is ^{99m}Tc, a radionuclide widely used i diagnostic nuclear medicine studies (6.01 hour half-life). To ensure that technetiu radiopharmaceuticals are readily available, many clinical nuclear medicine departments ru molybdenum-based technetium generators on site (in hospital radiopharmacies).

Fig. 2 Presentation of data for Molybdenum – *from article: Molybdenum*

The terms related to medical engineering were very important for "opening of the black box" of some medical equipment (as X-ray, Ultrasound, MRI, etc). Information was included also for specific electronic components used in the equipment – Fig.3, as well as their parameters, relevant for medical physics. A number of equipment were presented with educational engineering block diagrams – Fig.4. The General terms related to various monitors (and visualization in general) were often combined with similar terms from the other fields (mainly Imaging).

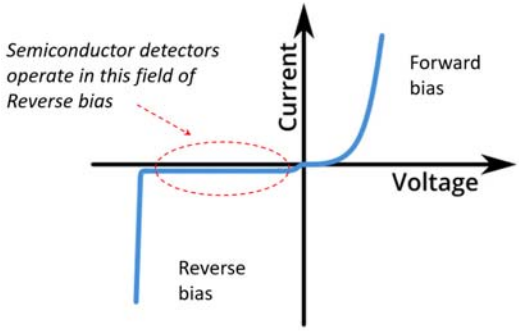


Fig.3 Semiconductor diode characteristic - *from article: Diode*

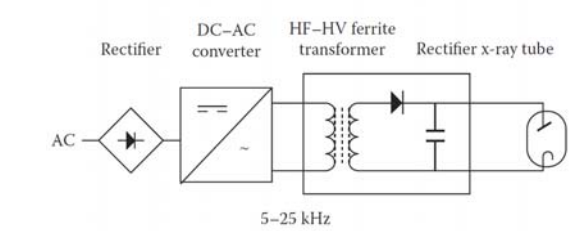


Fig.4 Block diagram of medium (high) frequency high-voltage x-ray generator - *from article: High-frequency generator*

A new area of the General terms, added in Edition II, was related to Medical Equipment management. The importance for including these was related to the fact that in small countries (especially in LMI countries) medical physicists often perform clinical engineering tasks. These terms covered the fields of Life cycle of equipment, Procurement, Servicing, etc. Fig. 5 gives an example for such terms.

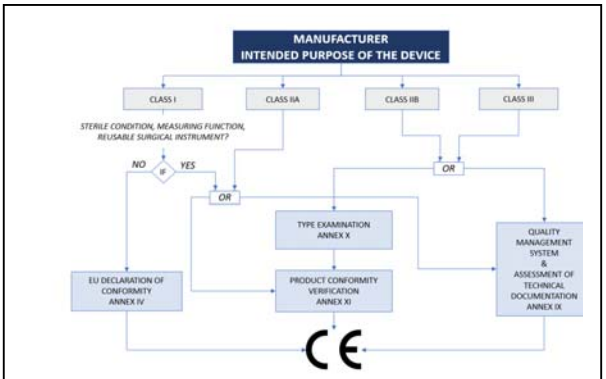


Fig.5 A diagram illustrating the necessary steps to obtain the CE mark for a medical device - *from article: Certificate of conformity*

Significant update of the information about PACS (Picture Archiving and Communication Systems) was updated (and new added) to cover this very dynamic new field of the profession. As with the engineering information, this area was often combined with terms from the Imaging fields. Fig. 6 gives a block diagram related to PACS.

Another General terms area covered information about various medical physics and related organisations and bodies, international projects/activities, professional development, etc. The existing data about IUPESM, IFMBE, IOMP, its Regional Organisations, etc was updated (these include various website addresses).

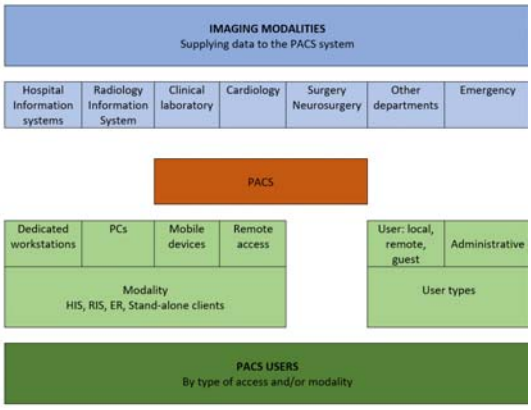


Fig.6 PACS - *from article: PACS*

The general terms also included brief articles about new areas entering medical as Artificial Intelligence, Data Mining, etc. Further references are shown for these areas.

III. CONCLUSION

The update of the field with General terms (some related to other fields) included about 110 new articles. These were managed by the Coordinators of the Working Group on General Terms: Slavik Tabakov, Magdalena Stoeva, Franco Milano, Ernesto Iadanza.

The update covered many new areas and included a lot of udates of existing articles. The Editorial Board shall be grateful to information from our colleagues about new methods and equipment to be included in the III Edition of the Encyclopaedia (possibly around 2031).

ACKNOWLEDGEMENTS

We gratefully acknowledge the contribution of so many colleagues from various countries to the update of the Encyclopaedia of Medical Physics – these are listed with Index 2 in the previous paper about the Encyclopaedia update [3] . Most active in II edition (General terms) were: E Iadanza, S Bergamasco, L Leogrande, L Pecchia, D Piaggio, S Tipnis, KP Lin, V Tabakova, C Caruana, S Mehta, P Sprawls.

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The Encyclopaedia of Medical Physics II Edition: The update of Diagnostic Radiology field

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Abstract: **The paper describes briefly the update of the Medical Physics Thesaurus of terms and the related update of Diagnostic Radiology field of the Encyclopaedia of Medical Physics II Edition (published 2021).**

Keywords *– Medical Physics Encyclopaedia, Medical Physics Education, Medical Physics resources.*

I. Introduction

The Encyclopaedia of Medical Physics development and its update are parts of a large project, which took over 20 years. The articles of the Encyclopaedia I Edition (published by CRC Press in 2013 [1]) were based on the Medical Physics Thesaurus of terms, developed in 2003 and updated in 2008. An additional minor update of the full Thesaurus was made in 2011. Thus, the Encyclopaedia Edition I included over 2800 articles explaining the foundation terms in medical physics. These were published by CRC Press (in paper) as a two-volume set and uploaded (together with the Scientific Dictionary of Medical Physics Terms in 32 languages) on the dedicated website www.emitel2.eu as a free reference and educational resource.

During the following 10 years materials for the Thesaurus update were collected and a major update was made in the period 2019-2020. This update included about 650 new terms. The Encyclopaedia II Edition is naturally listed alphabetically, but it has specific parts (fields), managed by different teams, as per the narrow specialty of the contributors. These fields are on Physics of: Diagnostic Radiology; Radiotherapy; Nuclear Medicine; Ultrasound Imaging; Magnetic Resonance Imaging; Radiation Protection; Non-ionising radiation protection; General terms (including Management). This new II Edition of the Encyclopaedia of Medical Physics was printed and published by CRC Press in 2021 [2]. The materials from the update were uploaded at the same website: www.emitel2.eu

II. Diagnostic radiology update

This paper describes briefly the nature of the update and the new terms in the filed of Diagnostic Radiology. The initial Thesaurus of Diagnostic Radiology terms (from 2003) included topics related to X-ray tubes and generators; X-ray equipment and stands; X-ray films; Computed Radiography systems; Various Digital detectors in Radiology; Computed Tomography; Specific physics-based methods in Diagnostic Radiology, etc.

During the 10 years period this field has undergone significant development, which was necessary to be included in the medical physics knowledge bank.

The Encyclopaedia Editorial Board decided to keep the historical parts in all fields, as a number of these included some important methods and scientific approaches, which can be used for future references. In Diagnostic Radiology such were, for example, the fields of Linear Tomography (now forming the foundation of the Digital Tomosynthesis), or Classical X-ray Generators (forming the foundation of the Medium frequency X-ray generators today), etc.

A number of the existing Encyclopaedia articles were updated to include new developments in Diagnostic Radiology – such as articles related to Multi Detector Computed Tomography and related fields, or X-ray imaging printing systems. A major update was necessary for the quickly developing field of Digital Detectors. These subjects also included updated and new articles associated with the Image assessment and Quality Control (QC) related to Digital Detectors.

New articles were included associated with new Test objects (phantoms) and parameters used for assessment of the detectors, as well as real images related to QC – example in Fig.1

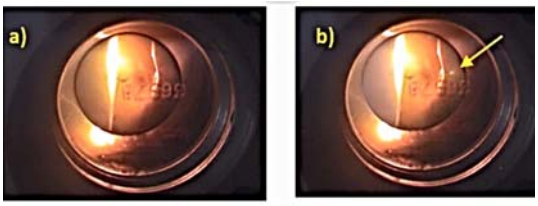

Fig.1 Arcing on X-ray tube photographed through the glass envelope: a) normal work of the X-ray tube (glowing-hot Anode on left and Cathode cup on right); b) small spark (arcing) appears close to Cathode - *from article: Arcing of X-ray tube*

Digital Tomosynthesis and various methods for Image Reconstruction were other fields, which have undergone significant update.

The Dual energy imaging description, as well as other Digital imaging methods, were explained in conjunction with the possible extraction of quantitative measurements used for diagnostic and QC purposes.

Special attention was given to fields, which are now forming important part of the educational process – such as visualisation of digital X-ray images (use of windowing and digital image filtration)- – example in Fig.2

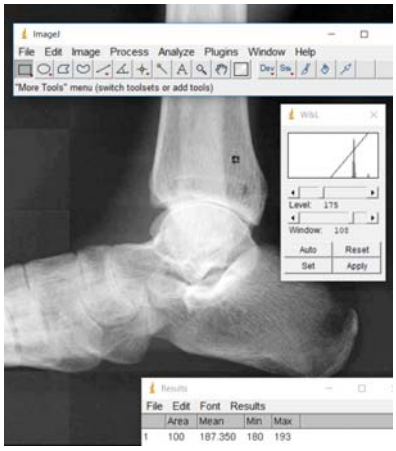

Fig.2 Measurement of pixel value (PV) in a square ROI of 100 pixels, showing mean pixel value of 187.35, while individua pixel values in the ROI vary from 180 to 193. Note that the pixel value does not change (compared with Fig.1), despite the visualization of the image with better contrast (resulting from the different window parameters - see the histogram) – *from article: Pixel value*

Optimisation was another field, where a number of new articles were included, as well as previous articles were updated. Such articles were often shown in sync with articles from other fields (e.g. Nuclear Medicine physics). In a similar way, for example, articles related to Monitors were shared with General terms (this is valid for all fields of the Encyclopaedia). Due to this reason in many parts of the Encyclopaedia the Related Articles (normally at the end of the text of most articles) were updated and enlarged.

The naming of the articles was kept in a way to present a cluster of these in close proximity inside the Encyclopaedia. All these articles were illustrated in a way to allow their use in the education process (information about real imaging systems was excluded in the articles) – example in Fig.3

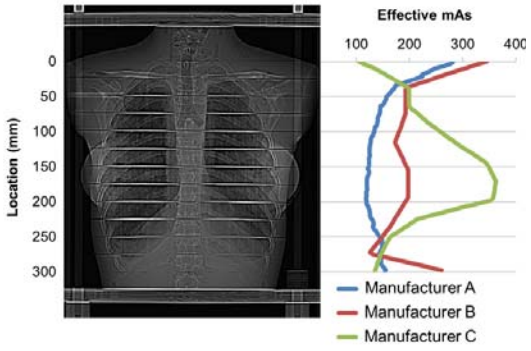

Fig 3 Comparison of the tube current when applying longitudinal modulation for different manufacturers – *from article: CT Optimization – longitudinal Modulation (z-axis)*

Short articles were included in connection with the use of DR methods outside their clinical application (such as Industry X-ray imaging).

Some very new emerging fields, such as Phase Contrast Imaging were included. In such case care was taken for the educational explanation of these and their good illustration. Many new diagrams and examples were presented to allow the reader to be introduced the field – example in Fig.4

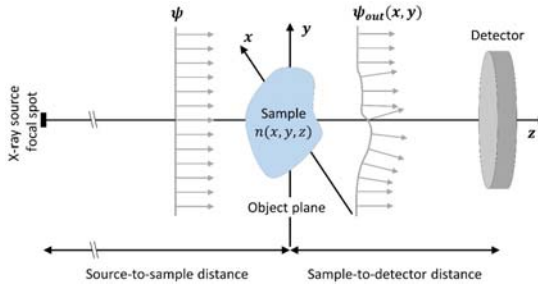

Fig 4 An X-ray beam, represented by the plane wave ψ, impinges on a sample described by the refractive index n(x,y,z). The wave ψout emerging from the object, is affected both in amplitude and in phase. Arrows represent the local direction of the X-ray wave – *from article: Phase Contrast Imaging*

As in the Edition I of the Encyclopaedia care was taken to include block diagrams and engineering information for the equipment, not taking it as a "black box". Fig. 5 gives an example.

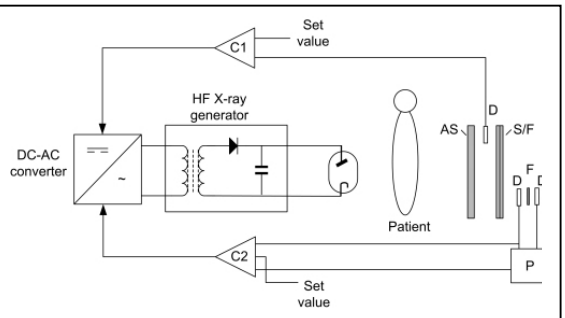

Fig 5. Block diagram showing two typical AEC types. C1 – used for chest radiography and C2 used for mammography – *from article: Automatic Exposure Control (AEC)*

III. CONCLUSION

The update of the DR field included about 150 new articles. These were managed by the Coordinators of the Working Group on Diagnostic Radiology (X-ray): Slavik Tabakov, Perry Sprawls, Paola Bregant.

The update covered most new areas of the dynamic field of Diagnostic Radiology. The Editorial Board shall be grateful to information from our colleagues about new methods and equipment to be included in the III Edition of the Encyclopaedia (possibly around 2031).

ACKNOWLEDGEMENTS

We gratefully acknowledge the contribution of so many colleagues from various countries to the update of the Encyclopaedia of Medical Physics – these are listed with Index 2 in the previous paper about the Encyclopaedia update [3] . Most active contributors in the II edition (DR filed) were: F Brun, S Pani, A Amin, G Havariyoun, L Brombal, C Anderson, H Delis, S Mehta, K Matsubara, K Ng.

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The Encyclopaedia of Medical Physics II Edition: The update of Nuclear Medicine and Ultrasound Sections

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Abstract: **The paper describes briefly the update of the Medical Physics Thesaurus of terms and the related update of Nuclear Medicine and Ultrasound fields of the Encyclopaedia of Medical Physics II Edition (published 2021).**

Keywords *– Medical Physics Encyclopaedia, Medical Physics Education, Medical Physics resources.*

I. Introduction

The Encyclopaedia of Medical Physics development and its update are parts of a large project, which took over 20 years. The articles of the Encyclopaedia I Edition (published by CRC Press in 2013 [1]) were based on the Medical Physics Thesaurus of terms, developed in 2003 and updated in 2008. An additional minor update of the full Thesaurus was made in 2011. Thus, the Encyclopaedia Edition I included over 2800 articles explaining the foundation terms in medical physics. These were published by CRC Press (in paper) as a two-volume set and uploaded (together with the Scientific Dictionary of Medical Physics Terms in 32 languages) on the dedicated website www.emitel2.eu as a free reference and educational resource.

During the following 10 years materials for the Thesaurus update were collected and a major update was made in the period 2019-2020. This update included about 650 new terms. The Encyclopaedia II Edition is naturally listed alphabetically, but it has specific parts (fields), managed by different teams, as per the narrow specialty of the contributors. These fields are on Physics of: Diagnostic Radiology; Radiotherapy; Nuclear Medicine; Ultrasound Imaging; Magnetic Resonance Imaging; Radiation Protection; Non-ionising radiation protection; General terms (including Management). This new II Edition of the Encyclopaedia of Medical Physics was printed and published by CRC Press in 2021 [2]. The materials from the update were uploaded at the same website: www.emitel2.eu.

This paper describes briefly the nature of the update and the new terms in the fields of Nuclear Medicine and Ultrasound. Since the initial publication of the Thesaurus of Diagnostic Radiology terms (from 2003) which included topics related to gamma cameras and ultrasound transducers, both the fields have undergone significant development, which was necessary to be included in the medical physics knowledge bank.

The Encyclopaedia Editorial Board decided to keep the historical parts in all fields, as a number of these included some important methods and scientific approaches, which can be used for future references.

II. Nuclear Medicine Update

In Nuclear Medicine, the retained topics included basic physics, radiation interaction, functioning of gamma cameras and SPECT systems etc. However, about 40 topics were updated and /or added to reflect the advances in the field since the last publication of this Encyclopaedia. For example, this edition includes an article on the revolutionary new total-body PET system as well articles on PET radiopharmaceuticals such as Rb-82, N-13 etc. Other new topics include nanoparticle and quantitative imaging.

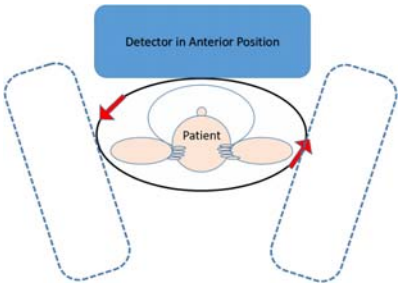
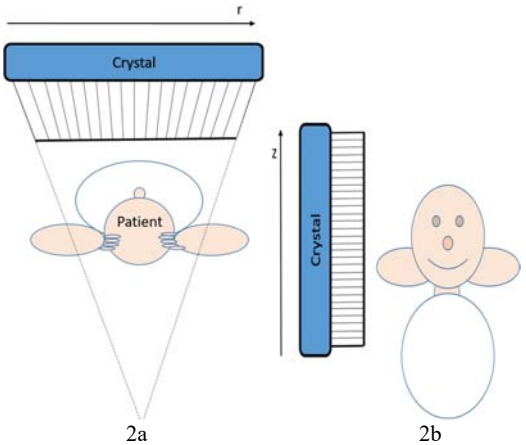


Fig.1 Body-contouring orbit in SPECT imaging

Some new articles emphasised the practical aspects of acquiring clinical data, such as the role of body-contouring orbits in SPECT data acquisition (see Fig. 1) and the role of fan-beam collimators for brain imaging (see Fig. 2).

Additionally, several new radiopharmaceuticals used in clinical nuclear medicine imaging such as pentetreotide, oxine, mebrofenin were added. These articles describe the practical use of these agents in clinical practice and should serve well to educate the reader about their specific roles in clinical imaging.

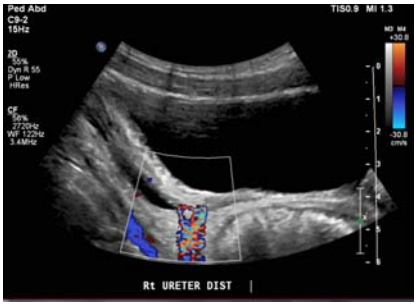


2a 2b

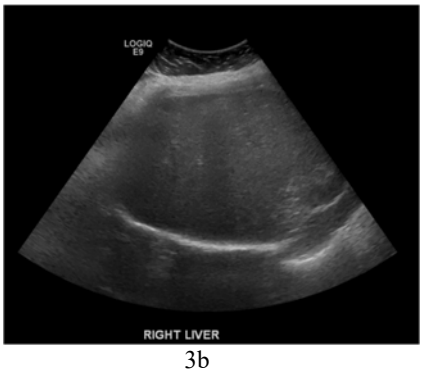
Fig.2 Fan beam collimator used in brain imaging. 2a - Fan Beam Collimator in the radial direction 2b - Fan Beam Collimator in the z direction

III. ULTRASOUND UPDATE

In Ultrasound imaging, the retained topics included basic physics of transducers, propagation of sound waves and interference phenomenon. About 35 topics were updated and /or added to reflect the advances in the field since the last publication of this Encyclopaedia. Of note is the inclusion of clinically relevant and common artefacts such as the twinkle artefact (Fig. 3) (common in doppler imaging) and speed displacement artefact. Other new topics include nanoparticle and quantitative imaging.



3a



3b

Fig 3a. Twinkle artefact in doppler imaging. 3b. Speed displacement artefact

Several other topics, such as 1.5 D transducer array, 4D ultrasound as well as techniques such as spatial compounding which are common to modern clinical ultrasound imaging were included.

Some new emerging fields, such as shear wave elastography are also included in this edition. While these techniques are were not commonly used about 10 years ago, advances in electronics, techniques and sophisticated transducers have now allowed them to be used for diagnostic imaging in many modern clinics. Fig. 4 is a clinical example of the shear wave elastography image.

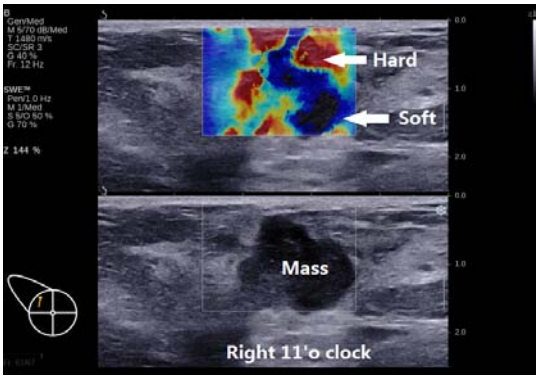


Fig 4 2-D shear wave elastography can help differentiate between hard and soft tissue

IV. CONCLUSION

The update of the nuclear medicine and ultrasound imaging includes about about 75 new articles. These were managed by the Coordinators of the Working Group on Nuclear Medicine and Ultrasound Imaging: Sameer Tipnis and Kwan Ng.

The update covered most new areas of these fields. The Editorial Board shall be grateful to information from our colleagues about new methods and equipment to be included in the III Edition of the Encyclopaedia (possibly around 2031).

ACKNOWLEDGEMENTS

We gratefully acknowledge the contribution of so many colleagues from various countries to the update of the Encyclopaedia of Medical Physics – these are listed with Index 2 in the previous paper about the Encyclopaedia update [3] . Most active in II edition (Nuclear Medicine and Ultrasound) were: C Bowen, J Taprogge, B Newman.

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The Encyclopedia of Medical Physics II Edition: The update of Magnetic Resonance Imaging Section

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Abstract: **The update describes briefly the update of the Medical Physics Thesaurus of terms in the field of Magnetic Resonance Imaging (MRI) within the Encyclopedia of Medical Physics II Edition (published 2021).**

Keywords *– Medical Physics Encyclopedia, Medical Physics Education, Medical Physics resources, Magnetic Resonance Imagin, MRI.*

I. INTRODUCTION

The Encyclopedia of Medical Physics (Edition I) was published by CRC Press in 2013 [1]. The terms included in the Encyclopedia were based on the Medical Physics Thesaurus of Terms developed in 2003 and updated in 2008. The first Edition of the Encyclopedia included articles explaining a wide range of terms in medical physics. Some of those terms were general terms, supporting a range of medical physics topics, and some were terms relating to specific imaging modalities. The Scientific Dictionary of Medical Physics Terms is a part of the Encyclopedia and it is available on the dedicated website www.emitel2.eu as a free reference and educational resource and is available in 32 languages.

In the years following the 2013 publication of the Encyclopedia, materials for the Thesaurus update were collected. A major update was made in 2019-2020. This update included about 650 new terms plus additional diagrams, tables and references. The Encyclopedia's second edition has specific fields managed by specialty teams. These fields are include; Diagnostic Radiology; Radiotherapy; Nuclear Medicine; Ultrasound Imaging; Magnetic Resonance Imaging; Radiation Protection; Non-ionising radiation protection; and General terms (including Management). The second edition of the Encyclopedia of Medical Physics was published by CRC Press in 2021 [2]. The updated terms materials from the update were uploaded to the website: www.emitel2.eu.

This article describes the update to the MRI section of the Thesaurus.

II. MAGNETCI RESOANCE IMAGING (MRI) UPDATE

Since the time of the last update of the encyclopedia, the field of MRI has expanded in both the number of scans conducted and the number of imaging devices in the field. Additionally, the range of techniques and the clinical applications of MRI has expanded. Finally, hardware improvements and developments have occurred. The update of the MRI section of the encyclopedia has attempted to reflect this change in the technology by adding terms and updating terms with new developments. In all, 30 terms were added or updated.

The MRI techniques added include new applications, which allow MRI to evaluate new biochemical pathways, for

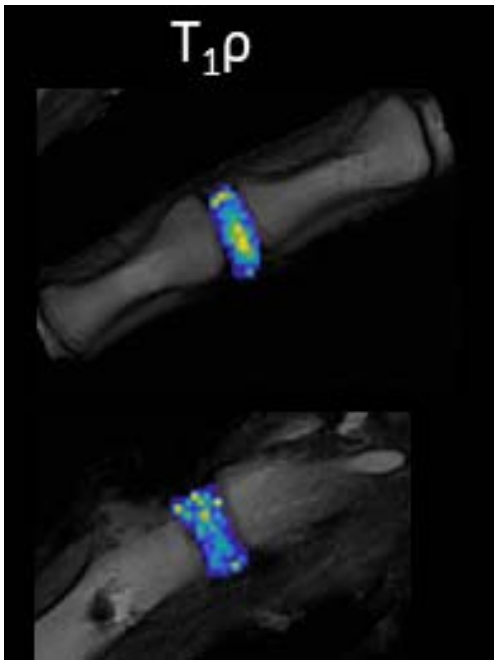

Figure 1. Images made with T1ρ contrast. An intervertebral disk (IVD) with normal collagen (top) and an IVD with deteriorated collagen.

example, Chemical Exchange Saturation Imaging (CEST), or new contrast mechanisms, for example, Susceptibility-Weighted imaging (SWI), Diffusion Kurtosis Imaging (DKI) and Ultrashort TE (UTE) imaging.

The next set of terms added related to the new technology, which has been applied to MRI in recent years to accelerate imaging and or improve image quality. Some of these methods/techniques have been borrowed from other fields and applied to MRI, for example Compressed Sensing. Other technical developments are specific to MRI, for example Multi Band/Simultaneous Multi-Slice (SMS) Imaging.

A group of terms was added relating to hardware developments, mostly in the expansion of the available field strengths now in use with MRI as shown in the new entries for Ultra-High field MRI (7 T and more), Multi-coil transmit, and Helium-free magnets.

Finally, some terms that are not necessarily new, but were missing from the original encyclopedia were added. Some terms from the original encyclopedia were updated to ensure clarity and consistency among terms.

The majority of new terms come with figures and references. The figure are a mix of illustrations that clarify the concepts presented or illustrate images acquired with the techniques. The references allow the reader to explore topics more in-depth. Illustration for T1-rhp (T1r) imaging show below as examples.

III. CONCLUSION

The update of the Magnetic Resonance Imaging field included about 35 new or updated articles. These were managed by the Coordinators of the Working Group on MRI Terms including John Oshinski, Renata Longo from Universita di Trieste (Trieste, Italy) and Antonio De Stefano from NHS Trust Portsmouth Hospital (Portsmouth, UK).

The update covered the addition of new technologies and application of MRI and revisions of existing articles.

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The Encyclopaedia of Medical Physics II Edition: The update of Radiation Protection Terms field

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Abstract: **The paper describes briefly the update of the Medical Physics Thesaurus of terms and the related update of Radiation Protection Terms field of the Encyclopaedia of Medical Physics II Edition (published 2021).**

Keywords *– Medical Physics Encyclopaedia, Medical Physics Education, Medical Physics resources, Radiation Protection, Ionizing Radiation, Non-Ionizing Radiation.*

I. Introduction

The Encyclopaedia of Medical Physics development and its update took over 20 years. The articles of the Encyclopaedia I Edition (published by CRC Press in 2013 [1]) were based on the Medical Physics Thesaurus of terms, developed in 2003 and updated in 2008. An additional minor update of the full Thesaurus was made in 2011. The Encyclopaedia Edition I included over 2800 articles explaining the foundation terms in medical physics. Some of those terms were General terms, supporting medical physics knowledge. Edition I was published by CRC Press (in paper) as a two-volume set and uploaded (together with the Scientific Dictionary of Medical Physics Terms in 32 languages) on the dedicated website www.emitel2.eu as a free reference and educational resource.

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II. Radiation protection terms update

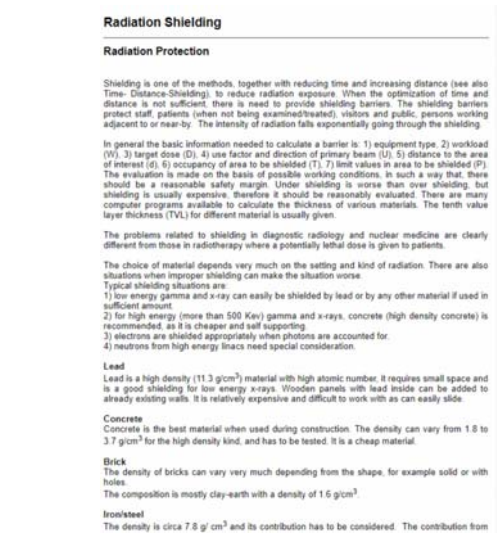
This paper describes briefly the nature of the update and the new Radiation Protection terms of the Encyclopaedia. The initial Thesaurus of Diagnostic Radiology terms (from 2003) included Radiation Protection terms related to various aspects of healthcare and medical physics, both in the field of ionizing and non-ionizing radiation.

Most of the Radiation Protection terms from ionizing radiation were covered in Edition I. Some additional terms with focus on non-ionizing radiation and organizations aspects related to radiation protection were added in Edition II.

A key aspect related to the Radiation Protection group of terms is that quite ofthen they are shared between two or more groups which proves their importance for medical physics and healthcare.

Fig. 1 presents the standard style of such term. In most of the cases they inlude primarily text information, certain numbers and formulas, and of external references.

The Ionizing Radiation terms included topics covering the various aspects of application of ionizing radiation in healthcare – diagnostics and treatment, as well as medical physicists responsibility. Special attention is paid to the protection of patients and staff. References to international gidance are also included in the form of links and further reading.



Radiation Shielding

Radiation Protection

Shielding is one of the methods, together with reducing time and increasing distance (see also Time- Distance-Shielding), to reduce radiation exposure. When the optimization of time and distance is not sufficient, there is need to provide shielding barriers. The shielding barriers protect staff, patients (when not being examined/treated), visitors and public, persons working adjacent to or near-by. The intensity of radiation falls exponentially going through the shielding.

In general the basic information needed to calculate a barrier is: 1) equipment type, 2) workload (W), 3) target dose (D), 4) use factor and direction of primary beam (U), 5) distance to the area of interest (d), 6) occupancy of area to be shielded (T), 7) limit values in area to be shielded (P). The evaluation is made on the basis of possible working conditions, in such a way that, there should be a reasonable safety margin. Under shielding is worse than over shielding, but shielding is usually expensive, therefore it should be reasonably evaluated. There are many computer programs available to calculate the thickness of various materials. The tenth value layer thickness (TVL) for different material is usually given.

The problems related to shielding in diagnostic radiology and nuclear medicine are clearly different from those in radiotherapy where a potentially lethal dose is given to patients.

The choice of material depends very much on the setting and kind of radiation. There are also situations when improper shielding can make the situation worse.
Typical shielding situations are:
1) low energy gamma and x-ray can easily be shielded by lead or by any other material if used in sufficient amount.
2) for high energy (more than 500 Kev) gamma and x-rays, concrete (high density concrete) is recommended, as it is cheaper and self supporting.
3) electrons are shielded appropriately when photons are accounted for.
4) neutrons from high energy linacs need special consideration.

Lead
Lead is a high density (11.3 g/cm^3) material with high atomic number. It requires small space and is a good shielding for low energy x-rays. Wooden panels with lead inside can be added to already existing walls. It is relatively expensive and difficult to work with as can easily slide.

Concrete
Concrete is the best material when used during construction. The density can vary from 1.8 to 3.7 g/cm^3 for the high density kind, and has to be tested. It is a cheap material.

Brick
The density of bricks can vary very much depending from the shape, for example solid or with holes.
The composition is mostly clay-earth with a density of 1.6 g/cm^3.

Iron/steel
The density is circa 7.8 g/ cm^3 and its contribution has to be considered. The contribution from

Fig.1 Common layout of a Radiation protection term

The Non-Ionizing Radiation terms section was expanded in the second edition of the encyclopedia to cover some of the most recent aspects related to the appliction of non-ionizing radiation in medical diagnostics and treatment. Many images, tables and diagrams are included with educational purpose (Fig.2). And again medical physicists responsibility, protection of patients and staff are on focus. References to international gidance are also included in the form of links and further reading.

The terms related to the organizational aspcts of radiation protection promarily refer to organizations working in the field, gidance and legislation, events.

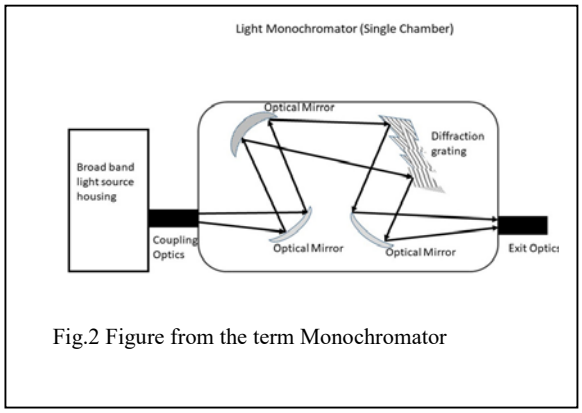


Fig.2 Figure from the term Monochromator

III. CONCLUSION

The update of the field with Radiation Protection terms (some related to other fields) included about 130 new articles. These were managed by the Coordinators of the Working Group on Radiation Protection Terms: Magdalena Stoeva and Jim Thurston. Non-ionising Radiation Safety Working Group Coordinators were: Fiammetta Fedele and Elizabeth Chaloner.

The update covered the contemporary aspects of radiation protection and included a lot of updates of existing articles. The Editorial Board shall be grateful to information from our colleagues about new methods and equipment to be included in the III Edition of the Encyclopaedia (expected around 2031).

ACKNOWLEDGEMENTS

We gratefully acknowledge the contribution of so many colleagues from various countries to the update of the Encyclopaedia of Medical Physics – these are listed with Index 2 in the previous paper about the Encyclopaedia update [3] . Most active in II edition (Radiation Protection) were: J Thurston, A Amin, F Milano, P Bregant, S Mehta, S Tabakov, K Matsubara.

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The Encyclopaedia of Medical Physics II Edition: Radiotherapy Update

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Abstract: **The paper briefly outlines the Radiotherapy update of the Encyclopaedia of Medical Physics II Edition (published 2021).**

Keywords *– Medical Physics Encyclopaedia, Medical Physics Education, Medical Physics resources, Radiotherapy.*

I. Introduction

The Encyclopaedia of Medical Physics aimed to collect, in our opinion for the first time, terms that contribute to a very extensive (if not total) knowledge of the most important applications of Physics in Medicine. The articles were originally based on a Medical Physics Thesaurus of terms, developed in 2003 and updated in 2008 and 2011. Thus, Edition I of the Encyclopaedia included over 2800 articles explaining the foundation terms in medical physics. These were published by CRC Press (in paper, in 2013 [1]) as a two-volume set and uploaded (together with the Scientific Dictionary of Medical Physics Terms in 32 languages) on the dedicated website www.emitel2.eu as a free reference and educational resource.

Over the period 2019-2020 our editorial sub-group made a major update of radiotherapy entries within both the Thesaurus and Encyclopaedia. This update included almost 200 new or substantially modified radiotherapy articles.

II. Radiotherapy update

Over the past decade, in particular, advances in: proton and ion therapy, radiobiology, dosimetry, dose-calculation, imaging / image registration, and equipment design have shaped our field. We felt it essential to add articles on these topics, expanding the Encyclopaedia's bank of radiotherapy knowledge.

We paid special attention to proton and ion therapy, including over 30 new articles in this area. These articles include: 'Pencil Beam Scanning (PBS)', 'Spot spacing' (see Figure 1), 'Multi-field optimisation' (see Figure 2), 'Proton Arc Therapy' and 'Single room particle therapy systems', 'Proton radiography' and 'Proton CT (pCT)'.

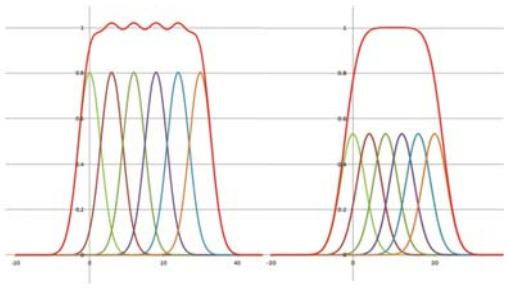

Fig.1 In scanned proton therapy, in order to produce a homogenous dose distribution, individual spots are delivered at a given distance from each other. The distance between each spot and the spot size will affect the homogeneity. If the spots are too far apart, the distribution becomes wavy. If spots are placed closer together, the weighting of each spot gets smaller. There is a limit to how low the weight of each spot can be, and below a certain limit they are undeliverable so the spacing must balance these two factors – *extracted from the 'Spot spacing' article*

The Encyclopaedia's radiobiology provision was expanded with new articles such as 'Track-structure' (see Figure 3), 'Double Strand Breaks', the 'Nonhomologous end-joining Repair Pathway', 'Hypoxia' and 'FLASH'.

For dosimetry, we added articles on topics such as 'Dose-to-medium calculations', 'Dose-to-water calculations',

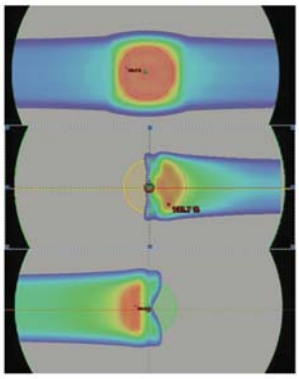

Fig.2 Multi-field optimisation is an optimisation technique which allows different scanned proton beams to cover different sections of the target. This technique is especially useful for a situation where the treatment volume is partially obstructed by an Organ at Risk (OAR) – *extracted from the 'Multi-field Optimisation' article*

Fig 3 The track structure along the path of a charged particle can be split into two regions, the infra-track and ultra-track (as shown above). The infra-track region encompasses the region where the electric field of the charged particle is sufficient to directly cause ionisations. The ultra-track region encompasses the infra-track and constitutes the region in which ionisations are caused by secondary electrons from the infra-track ionisations. The size of the ultra-track depends on the maximum energy of the secondary electrons produced by the charged particle – *extracted from the 'Track Structure' article*

'Faraday Cup', 'Alanine' and 'Metal oxide semiconductor field-effect (MOSFET) transistors'.

Considering imaging and registration, the encyclopaedia was updated to reflect advances in MR-guided radiotherapy with new articles such as: 'MR-linac', 'MRI-guided radiotherapy (MRIgRT)', 'MR-only treatment planning', 'Pseudo CT' and 'Deformable Image Registration (DIR)' (see Figure 4).

We also took care to reflect advances in treatment planning with articles such as 'Dose Painting', 'Multi-criteria Optimisation (MCO)' and 'Pareto Surface'.

Finally, we expanded the encyclopaedia to cover clinical terms relevant to radiotherapy, such as: 'Clinical Trial Endpoints', 'Patient Reported Outcome Measures

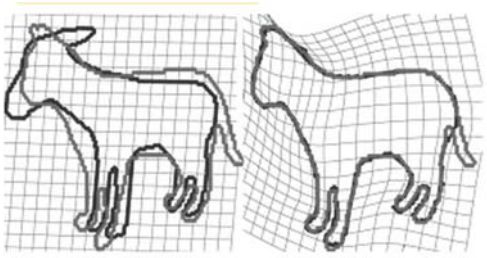

Fig 4 Deformable image registration (DIR) is the process of registering an image data set to a reference image data set by applying an elastic deformation to minimise the difference between the two. This allows comparison or integration of data which is obtained from two different measurements – *from the 'Deformable Image Registration (DIR)' article*

(PROMs)', 'Acute Morbidity', 'Long Term Morbidity' and 'Quality-adjusted life years (QALYs)' (see Figure 5).

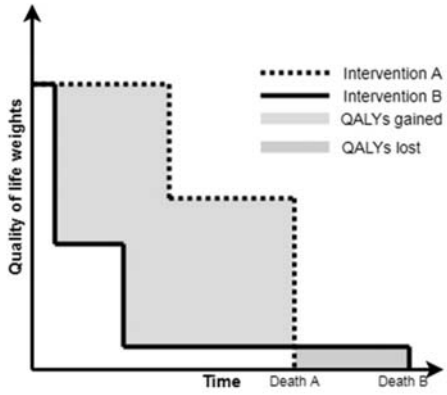

Fig 5 In order to have a systematic way to appraise the benefit of treatment options, many health organisations use quality-adjusted life years (QALYs) as the comparison metric. In contrast to metrics of quantity of life (e.g. overall survival), a QALY is a measure of the state of health of a person or group which incorporates quality of life. One QALY is equal to one year of life in perfect health. The condition of perfect health is measured in terms of the person's ability to carry out normal daily activities, freedom of physical pain and mental disturbance. This is calculated by estimating the years of life remaining for a patient after treatment or intervention and weighting each year on a scale of 0 to 1 (0 = death, 1 = best possible health state). – *extracted from the 'Quality-adjusted life years (QALYs)' article.*

III. CONCLUSION

The update of the radiotherapy field included almost 200 new articles. These were managed by the Coordinators of the Working Group on Radiotherapy – Franco Milano, Eva Bezak and Tracy Underwood.

The update covered exciting new developments in radiotherapy and expanded the bank of terms related to clinical practice. The Editorial Board would welcome suggestions from colleagues regarding new radiotherapy methods and equipment that should be included in the III Edition of the Encyclopaedia (possibly around 2031).

ACKNOWLEDGEMENTS

We gratefully acknowledge the contribution of so many colleagues from various countries in producing both Edition 1 and Edition II of the Encyclopaedia of Medical Physics – these are listed with Index 2 in the previous paper about the Encyclopaedia update [3] .

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PROFESSIONAL TOPICS

IAEA TRAINING RESOURCES ON RADIATION PROTECTION IN DENTAL RADIOLOGY

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Abstract— **The International Action Plan for the Radiological Protection of Patients and the Bonn Call for Action emphasised the need to strengthen radiation protection education and training of health professionals, tailored to the practical needs of different audiences and taking into account their specialties and field of work. The IAEA standard training package and the e-learning on Radiation Protection in Dental Radiology are the newest resources to support education and training in radiation protection of dental professionals. They have been developed in collaboration with several professional bodies. The standard training package of 12 lectures has been designed to support trainers providing radiation protection education and training of different professionals involved in dental radiology – dentists and other dental staff, dental radiation technologists, medical physicists. The e-learning structured in 9 modules is adapted to the self-learning by dentists and other dental professional staff. These training resources are freely available from the Radiation Protection of Patients (RPOP) webpage of the IAEA, www.iaea.org/resources/rpop.**

Keywords— **Radiation protection, Education and training, Training material, e-learning, Dental radiology**

I. IAEA'S STRATEGIC APPROACH FOR SUSTAINABLE TRAINING ON RADIATION PROTECTION IN MEDICINE

International Atomic Energy Agency (IAEA) established in 2002 an International Action Plan for the Radiological Protection of Patients in cooperation with international organizations and professional bodies, with the objective to make progress in patient safety through variety of actions [1]. The education and training of healthcare professionals have been recognized as essential for achieving this goal. Actions to strengthen education and training included the development of standard syllabi and packages for training of health professionals and to train the trainers involved in national training programmes using this material. It has been stressed that the training programmes and training material must be tailored to the practical needs of different audiences, taking into account their specialties and field of work, and to make the material available in the official languages of the United Nations.

The Bonn Call-for-Action, a joint Position Statement issued by the IAEA and World Health Organization (WHO) in 2012, further emphasized in the Action 4 the need to strengthen radiation protection education and training of health professionals, by a) prioritizing radiation protection education and training for health professionals globally, targeting professionals using radiation in all medical and dental areas; b) further developing the use of newer platforms such as specific training applications on the Internet for reaching larger groups for training purposes; c) integrating radiation protection into the curricula of medical and dental schools, ensuring the establishment of a core competency in these areas; d) strengthening collaboration in relation to education and training among education providers in health care settings with limited infrastructure as well as among these providers and international organizations and professional societies; e) paying particular attention to the training of health professionals in situations of implementing new technology [2].

Since 2002 and following the establishment of the Radiation Protection of Patients Unit of the IAEA, different resources have been developed to support education, training and risk communication, all freely available from the dedicated website on Radiation Protection of Patients (RPOP), https://www.iaea.org/resources/rpop. Currently, 14 training packages are available in different areas and specific applications, all developed with the involvement of internationally recognized experts and in close collaboration with appropriate international and professional organizations. These materials contain PowerPoint slides, available in English, Spanish and Russian for free download from the training page of the website [3]. With the purpose to train trainers and receive feedback, the IAEA organizes every year a number of training events based on these standard syllabi and materials. The IAEA packages have become a leading training resource on radiation protection in medicine used worldwide. The material is of particular importance for less resourced countries, where it is translated into local languages, adapted and used by trainers for organizing trainings at national or local level.

The IAEA regularly updates the existing and develops new training material to address the new developments and reach wider audience of health professionals. This includes both new topical areas and new training approaches such as e-learnings for direct self-learning and webinars.

The above-mentioned actions support the implementation in the IAEA Member States of the requirements of the International Basic Safety Standard, GSR Part 3, and the recommendations of the associated Safety Guide SSG-46 related to medical uses of ionizing radiation [4,5].

II. IAEA RESOURCES FOR RADIATION PROTECTION IN DENTAL RADIOLOGY

X-ray imaging is extensively used in dentistry to diagnose, plan and monitor treatments and to follow-up pathoses.

According to the latest report of the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR), 1.1 billion dental examinations are performed globally every year, that is an increase of 130% from the 480 million examinations per annum estimated in the UNSCEAR 2008 report [6]. Dental examinations account for approximately 26 % of all diagnostic radiological examinations, with the annual frequency estimated to 151 dental examinations per 1,000 population globally, varying from 21 per 1,000 population in low-income countries, to 561 per 1,000 in high-income countries [6].

Dentistry is an independent healthcare specialty. Dental X-ray equipment is often owned by dentists, who refer patients for X-ray procedures performed by themselves. Therefore, dentists have a responsibility for justification of medical exposure and for optimization of radiation protection of patients, and they would be more effective if they have a good understanding of radiation risks and radiation protection and safety. The RPOP website offers answers to frequently asked questions and other information for dental professionals and patients [3].

Specific guidance on radiation protection in dental radiology is provided in the new IAEA Safety Report developed in cooperation with the FDI World Dental Federation, the Image Gently Alliance, the International Association of DentoMaxilloFacial Radiology (IADMFR) and the International Organization for Medical Physics (IOMP) [7].

Radiation protection education and training helps dentist fulfil minimize potential misuses of self-referral and perform exams safely. Although formal processes to require such education and training under a radiation protection and safety framework might be difficult to put in place, a more general approach is recommended for promoting education and training in radiation protection and safety as part of the general university degree curriculum, especially in the course of dental imaging, and/or as part of the corresponding specialty postgraduate education and training programme [5, 7].

The IAEA standard training package on Radiation Protection in Dental Radiology and the e-learning on the same topic are the newest materials to support education and training in this area.

III. STANDARD TRAINING MATERIAL

The standard training syllabus and material on Radiation Protection in Dental Radiology have been designed to support trainers providing radiation protection education and training of different professionals involved in dental radiology – dentists and other dental staff, dental radiation technologists, medical physicists. It can be used by lecturers who teach to students in dental schools and schools for medical radiation technologists. The material is useful for regulators and inspectors who want to enhance their knowledge on specific aspects of radiation protection in dental radiology.

The training is structured in 12 lectures, the topics of which is presented in Table 1.

For each topic, a slide set in Power point (Microsoft Office) is provided, structured in the following sections: Educational Objectives; Overview; Content; References.

Table 1 Training modules of the standard training material

Lecture #	Topic
1	General Principles of Radiation Protection
2	Special Considerations for Radiation Protection in Children
3	X-ray Production and Interaction: Image Formation and Image Quality
4	General Principles of Film and Digital Radiography
5	Fundamentals of Intraoral Radiography
6	Fundamentals of Panoramic Radiography
7	Fundamentals of Extraoral Projectional Radiography
8	Fundamentals of CT and CBCT
9	Justification and Appropriate Use of Dental Radiology
10	Quality Assurance in Dental Radiology
11	Optimization of Protection of Patients in Dental Radiology
12	Protection of Workers and Public in Dental Radiology

Lecture 1, General Principles of Radiation Protection, provides general introduction to ionization radiation, its properties and interaction with matter, dose quantities and their use, different types of radiation risks and factors, and introduces the general principles of radiation protection and their application in the International Basic Safety Standards. At the end of this module, the participant should be able to:

- Understand the importance of adhering to the principles of radiation protection in dentistry;
- Understand the properties of ionizing radiation (especially X-rays), and their effects on living tissue;
- Distinguish between stochastic effects and tissue reactions;
- Distinguish between absorbed dose, equivalent dose and effective dose;
- Understand the linear-non-threshold model and its implication for radiation protection;
- Understand the dose-risk relation of stochastic effects, and the effect of age and gender;
- Understand the principles of justification, optimization of protection, and application of dose limits.

Lecture 2, Special Considerations for Radiation Protection in Children**,** explains why extra care is needed when exposing children to radiation and what are the approaches to protect pediatric patients when performing radiography.

The motivation for this separate module is the high frequency of dental radiographic examinations in children, used in orthodontic treatment, trauma, cleft palate, developmental disorders, tumors and other clinical situation. After learning this module, student should be able to:

- Understand the need for special considerations for radiation protection in children;
- Understand why radiation doses for children are higher than for adults (unless exposure parameters are adapted appropriately);
- Understand why radiation risks for children are higher.

Lecture 3, X-ray Production and Interaction: Image Formation and Image Quality, provides a general introduction to the basic physics of X-ray beam, components of an X-ray tube and key exposure parameters. Interactions of X rays with matter are discussed, as well as the principles of image formation and image quality parameters. After studying this topic, participants should be able to:

- Understand the function of the different components of the X-ray tube;
- Understand the effect of kV, mAs and filtration on the quantity and energy of X-rays;
- Understand the effect of geometric exposure parameters;
- Understand the basic principles of image formation in X-ray imaging;
- Be familiar with the different essential image quality characteristics.

The next five modules focus on important features of the four different X-ray imaging modalities used in dentistry and practical aspects of their proper use for obtaining diagnostic images at reasonably low dose to patients. First, principles of image formation in film and digital radiography are discussed, and further detailing in the next modules specific applications in intraoral, extraoral radiography and CT, including CBCT.

Lecture 4, General Principles of Film and Digital Radiography, has the following learning objectives:

- Understand the principle of image formation in film and digital radiography, and fundamental differences between them;
- Be familiar with image manipulation methods in digital radiography.

Lecture 5, Fundamentals of Intraoral Radiography, has the following objectives:

- Be familiar with the different types of intraoral radiography;
- Understand the difference between (film-based and digital) image receptors used in intraoral radiography;
- Recognize and avoid faulty radiographs (position, under/overexposure, film handling and development, etc.);
- Know when and how to use handheld intraoral radiography machines.

Lecture 6, Fundamentals of Panoramic Radiography, aims to make participants:

- Understand the general principles of image acquisition in panoramic radiography;
- Recognize and identify causes of image distortion and artefacts;
- Able to apply proper patient positioning and alignment;
- Recognize image aberration due to mispositioning.

Lecture 7, Fundamentals of Extraoral Projectional Radiography, has the objective to make participant differentiate between lateral cephalometric, posteroanterior cephalometric, submentovertex, occipito-mental and occipito-frontal projections, in terms of positioning of patient and image receptor and respective clinical applications.

Lecture 8, Fundamentals of CT and CBCT, focuses on the relatively higher dose modalities used in dentistry, which proper use requires the users to be familiar with image formation and factors influencing dose and image quality. Through this module, the student will:

- Understand the general principle of image acquisition and reconstruction in CT and CBCT;
- Understand fundamental similarities and differences between MDCT and CBCT, and how this affects their respective clinical application;
- Be familiar with CT image manipulation and visualization methods;
- Understand the cause and effect of various types of CT artefacts;
- Understand the issues related to density estimations in CBCT.

The next four modules discuss specific aspects of radiation protection of patients, workers and public in dental radiology, based on the international recommendations and good practice.

Lecture 9, Justification and Appropriate Use of Dental Radiology, focuses on one of the two main principles of radiation protection, and approaches for selection of an appropriate imaging technique in a given situation, thus "doing more good than harm". As justification process needs to take into account patient dose, typical doses from different dental modalities are given, discussing also their variation and proper communication (Table 2).

Referral guidelines for imaging, often called "selection criteria" in dentistry, are considered to be an important tool for justification, and examples of such guidelines are given for different clinical applications, such as caries diagnosis, orthodontics, periodontics, endodontics, implant planning, tooth extraction, other surgery, temporomandibular joint imaging.

Table 2 Typical doses from dental imaging procedures

Dental procedure	Typical effective dose	Equivalent period of exposure to natural radiation
Intraoral radiograph	0.3-21.6 µSv	1 h – 3 d
Panoramic radiograph	2.7-38 µSv	10 h – 6 d
Lateral cephalometric radiograph	2.2-14 µSv	8 h – 2 d
CBCT	11-1025 µSv (generally <300 µSv)	1.5 d – 5 m (generally <1.5 m)
CT (mandible)	250-1410 µSv	1 – 7 m
CT (mandible & maxilla)	430-860 µSv	2 – >4 m

At the end of this module, imaging in pregnancy is discussed, and also how to communicate information about risks and benefits to patients.

After studying this module, participants will be able to:

- Understand the general principles regarding the use of radiation in medicine;
- Judge the appropriateness of using 2D and 3D imaging techniques for a given patient;
- Can judge the current referral criteria for CBCT for various clinical applications.

Lecture 10, Quality Assurance in Dental Radiology, presents an overview of general principles of QA and QC, QC test protocols per modality (an example shown on Fig. 1). Dosimetry and dose monitoring and assessment of clinical image quality assessment are also included. Although not aiming to provide detailed guidance on QA and QC tests, it provides reference to published protocols stressing on the leading role of medical physicists.

The learning objectives of this module are to:

- Understand the general aspects of QA in radiology, and particular QA and QC aspects in dental radiology;
- Able to perform QC tests for different dental radiographic equipment (if responsible for such task).

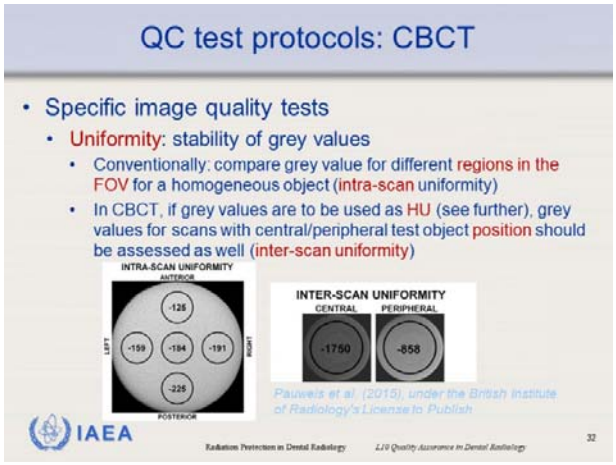

Fig. 1 An example from the training package, Module 10

Lecture 11, Optimization of Protection of Patients in Dental Radiology, discusses strategies to optimize imaging procedures for achieving diagnostic image at appropriate radiation dose to patient from different dental radiographic modalities, and the parameters affecting dose. The use of diagnostic reference levels (DRLs) and examples of available DRLs are presented. At the end of this module, the proper use of patient shielding is discussed. The learning objectives include:

- Be familiar with the equipment and patient factors affecting patient dose;
- Be able to apply various optimization strategies, and estimate the amount of dose that can be saved in a given exposure condition;
- Be able to judge the potential benefit and drawback of using patient shielding in a given situation.

Lecture 12, Protection of Workers and Public in Dental Radiology, presents regulatory and practical aspects of protection of workers when performing dental X-rays, and approaches to ensure protection of public, room design and shielding requirements. At the end of this module, the participants will be able to:

- Understands the scatter distribution for different radiographic modalities;
- Identify the safest position of the operator and public relative to the patient and x-ray tube;
- Understand the need for adequate distance and shielding and be able to apply these principles in clinical practice.

The training package was developed in several steps. First, with the involvement of invited experts, the syllabus and learning objectives were designed, followed by the development of Power point slides to deliver the content.

At the second step, experts from the IAEA, WHO and several international professional bodies representing the professional groups involved in dentistry were invited to review the draft material and provide their feedback. Representatives of all these organizations were invited to Vienna for a consultancy meeting, during which an updated version of the material was prepared, followed by review and approval by the organizations involved in the development of the material: WHO, FDI World Dental Federation, IADMFR, IOMP and Image Gently Alliance.

The approved training material was posted at the RPOP training webpage in 2017. During 2018, the training material was translated into Spanish. The training package on Radiation Protection in Dental Radiology can be downloaded from this link: https://www.iaea.org/resources/rpop/resources/training-material#12.

To promote the new material, IAEA organized two webinars within the RPOP webinar series, with lecturers who were actively involved in developing the training package, on the following topics:

- Optimization of dental CBCT exposures: a practical guide, with Dr. Ruben Pauwels from Belgium;
- Justification of X-ray examinations in dentistry, with Prof. Keith Horner from United Kingdom.

These recorded lectures and two other recent webinars organized jointly with the IADMFR in 2020 are available for free viewing from the RPOP webinar page: https://www.iaea.org/resources/rpop/resources/webinars.

IV. E-LEARNING MATERIAL

The online course on Radiation Protection in Dental Radiology was provided in 2021. Its objective is to provide education in radiation protection for dentists and other dental professional staff. The format is adapted to the self-learning purposes by playing the role of a dental professional at The Family Dental Centre.

It contains nine modules summarised in Table 3, and the learning objectives presented below.

Table 3 Training modules of the e-learning course

Module #	Topic
1	'Rays and grays' 1: Understanding X rays
2	'Rays and grays' 2: What do we mean by radiation dose?
3	How it works: the technology of X ray imaging
4	Choosing the right X ray examination: basic principles
5	Choosing the right X ray examination: children and young people
6	Choosing the right X ray examination: adult patients
7	Optimization: keeping patient doses as low as diagnostically acceptable
8	Optimization: maintaining high quality in radiology
9	Protection of staff and the public who are not patients

Module 1, 'Rays and grays' 1: Understanding X rays, has the following learning outcomes:

- To be able to describe the nature of X rays;
- To explain how X rays are produced in a typical dental X ray set;
- To recognize how exposure settings change the quantity and quality of X rays;
- To outline what happens in X ray attenuation.

Module 2, 'Rays and grays' 2: What do we mean by radiation dose?, has the following learning outcomes:

- To understand how X rays interact with tissues at the cellular level;
- To recognize the difference between stochastic effects and tissue effects;
- To understand what is meant by absorbed dose and effective dose;
- To recall the typical doses seen in dental X ray examinations;
- To be able to explain risks from X rays to patients;
- To recall the fundamental principles of radiation protection.

Module 3, How it works: the technology of X ray imaging, has the following learning outcomes:

- To understand the general principles of image acquisition in intraoral and panoramic radiography;
- To understand the difference between film-based and digital image receptors used in dental radiography;
- To know when and how to use handheld intraoral radiography machines;
- To understand the general principle of image acquisition and reconstruction in cone beam CT.

Module 4, Choosing the right X ray examination: basic principles, has the following learning outcomes:

- To understand the general principles of image acquisition in intraoral and panoramic radiography;
- To understand the difference between film-based and digital image receptors used in dental radiography;
- To know when and how to use handheld intraoral radiography machines;
- To understand the general principle of image acquisition and reconstruction in cone beam CT.

Module 5, Choosing the right X ray examination: children and young people, has the following learning outcomes:

- To develop knowledge of appropriate use of dental imaging for patients who are children or young people in different clinical contexts;
- To recognize the special considerations in justification for imaging children or young people;
- To recognize that it is inappropriate to select any X ray examination without first knowing patient history and carrying out a clinical examination;
- To understand that choice of imaging must be based on the reason for patient attendance (symptoms, clinical signs and past history).

Module 6, Choosing the right X ray examination: adult patients, has the following learning outcomes:

- To develop knowledge of appropriate use of dental imaging for adult patients in different clinical contexts;
- To recognize that it is inappropriate to select any X ray examination without first knowing patient history and carrying out a clinical examination;
- To understand that choice of imaging must be based on the reason for patient attendance (symptoms, clinical signs and past history);
- To recognize that routine screening examinations lead to overuse of X rays.

Module 7, Optimization: keeping patient doses as low as diagnostically acceptable, has the following learning outcomes:

- To recognize what is meant by 'optimization';
- To recall the features of intraoral, panoramic and cephalometric radiographic equipment that affect dose to patients;
- To recall the features of cone beam CT equipment that affect dose to patients;
- To recall the features of image receptors that affect dose to patients;
- To recall the role of shielding in reducing dose to patients;
- To recognize the value of dose surveys and diagnostic reference levels in optimization;
- To apply optimization in practice.

Module 8, Optimization: maintaining high quality in radiology, has the following learning outcomes:

- To recognize what is meant by quality assurance (QA) and quality control (QC);
- To appraise clinical image quality and differentiate between causes of loss of quality;
- To recognize how to perform QA and QC for X ray equipment;
- To recognize how to perform QA and QC for film imaging;
- To recognize how to perform QA and QC for digital imaging.

Module 9, Protection of staff and the public who are not patients, has the following learning outcomes:

- To understand how to limit X ray exposure to staff and the public who are not patients;
- To understand the scatter distribution for different radiographic modalities;
- To recognize the importance of distance, working time and barriers to protection;
- To know whether personal dose monitoring is relevant;
- To apply this knowledge to everyday practice as a dentist.

The final quiz consists of 30 questions and the users need to answer at least 80% correctly in order to obtain a certificate of completion.

The e-learning is currently available in English and Spanish, and since its launching in 2021, more than 3700 people studied it and over 2500 completed the course successfully.

The e-learning on Radiation Protection in Dental Radiology can be accessed from this link: https://www.iaea.org/resources/rpop/resources/online-training#6.

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THE ASEAN DIAGNOSTIC REFERENCE LEVELS IN MEDICAL IMAGING

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Abstract— **The Association of South-East Asian Nations, ASEAN, was established on 8 August 1967 in Bangkok, Thailand, with the Bangkok Declaration signed by the Founders of ASEAN, namely Indonesia, Malaysia, Philippines, Singapore and Thailand. Brunei Darussalam, Viet Nam, Lao PDR and Myanmar, and Cambodia were later joined, making up what is today the ten Member States of ASEAN. In the year 2000 at the World Congress on Medical Physics and Bio-Medical Engineering, held in Chicago, Illinois USA, the South-East Asian Federation of Organizations for Medical Physics (SEAFOMP) was established. The Congress of SEAFOMP-SEACOMP, is annually organized at each member state. The congress program consists of John Cameron Memorial Lecture, medical physics in radiation oncology, nuclear medicine, and diagnostic radiology on research and development, education and clinical training of medical physicists and radiation protection of staff and patients. The concept of Diagnostic Reference Levels in Medical Imaging had been introduced to SEAFOMP in early 2021. In ASEAN, DRLs was firstly started in Philippines in 2005, Malaysia in 2007, Vietnam in 2014, Indonesia in 2015 and Thailand in 2017. Laos, Cambodia and Myanmar join SEAFOMP on DRLs in 2021.**

Keywords— **ASEAN, SEAFOMP, Medical Imaging, NDRLs, optimization.**

Figure 1: The map of ASEAN Region

I. INTRODUCTION

The concept of Diagnostic Reference Levels has been introduced by ICRP, IAEA BSS, radiation protection of patients, RPOP, with the purpose of the optimization of the patient dose together with the adequate image quality. Furthermore, the College on Medical Physics organized the course on DRLs at Abdus Salam International Centre of Theoretical Physics co-host by IAEA, to the large number of participants from different regions at Trieste Italy. Those activities encouraged the optimization of the patient dose through the survey from frequent examinations with high patient dose such as CT, interventional radiology and other medical imaging. Several ASEAN members have no opportunity to start the survey according to the lack of medical physicist in medical imaging such as Brunei, Cambodia, Lao PDR, Myanmar, Singapore and Vietnam. Under the support from SEAFOMP President, Dr. Freddy Haryanto, Dr. Anchali Krisanachinda, President of Thai Medical Physicist Society had invited three distinguished consultants in early 2021 to initiate the national diagnostic reference levels and the regional diagnostic reference levels for ASEAN. Those are:

1. Dr. Donald McLean, Former IAEA Technical Officer, Human Health, IAEA Expert to Thailand, currently, Diagnostic Radiology Medical Physicist at Canberra Hospital, Australia.
2. Dr. Noriah Jamal from Malaysia, Former National Liaison Officer of Malaysia and Lead Country Coordinator on the IAEA Medical Physics Project
3. Professor Kosuke Matsubara, Kanazawa University, Japan, J-RIME committee on the establishment of Japan DRLs 2020.

The Consultants had paid the most important roles on giving lectures on DRLs, exchange their evidence and experience in establishing DRLs in their own countries and others, draft the excel file on the survey of patient data

related to dose quantities, patient indications and examinations, plan for national and regional DRLs activities, encourage the Member States to get along with the activities and contribute to the virtual meetings in 2021.

II. VIRTUAL MEETINGS

According the COVID 19 pandemic, the virtual meeting had been planned and organized for SEAFOMP members. Four virtual meetings had been hosted by Chulalongkorn University (Zoom Chula), Bangkok Thailand in 2021 as following:

***First*:** On May 27, 2021, three lectures titled
- *Concept of NDRLs, RDLs - NJ*
- *Templates for the regional data collection -DM*
- *Experience on Japan DRLs2020 -KM*
- *Country Reports-Member States*

Second: On July 21, 2121, three lectures titled
- *Survey on General Radiography -NJ*
- *Survey on Interventional Radiology -DM*
- *Survey on Computed Tomography -KM*
  - *Country Reports Member States*

Third: On September 9, 2021, four lectures titled
- *Survey on* Dental *Radiography -DM*
- *Survey on Fluoroscopy -KM*
- *Survey on Mammography -NJ*
- *Survey on Nuclear Medicine -AK*

Fourth: On December 23, 2021, three lectures titled
- *The 2015 Japan DRLs establishment -KM*
- *Proposed 2022 ADRLs Milestones -NJ*
- *Australian – UK DRLs -DM*
- *Country Reports on Milestone in 2022*

III. COUNTRY REPORTS

Cambodia

Mr. Ath Vannyat proposed the survey of CT, Interventional Radiology (IR) and General X-ray examinations. In the year 2018, the number of CT system in Cambodia was 58. Dose quantity is $CTDI_{vol}$ (mGy) and DLP (mGy.cm). The CT examination includes brain, cervical spine, thorax and lumbar spine. The number of general X-ray system was 520, the dose quantity is KAP ($mGy.cm^2$). X-ray examination includes Abdomen AP, Cervical Spine AP, LAT, Chest PA, LAT, Thoracic spine AP, LAT, Lumbo-sacral AP, LAT, Pelvis AP, Skull AP/PA, LAT. The number of X-ray systems in IR was 20. The patient size is weight (kg) and height (cm), dose quantity is KAP ($mGy.cm^2$), and the examination includes biliary intervention and facet joint intervention.

Indonesia

In May 2021, Indonesia established the national DRL (Indonesian Diagnostic Reference Level, IDRL) resulting from a series of coordination meetings among stakeholders over online-based data collection/survey initiated in 2015 by the Indonesian Nuclear Energy Regulatory Agency (BAPETEN). The established IDRL has been published and accessible at https://idrl.bapeten.go.id/index.php/site/idrl. Currently, it provides national DRL values for general radiography and CT for a typical adult patient. The indicated dose quantities were Entrance Surface Air Kerma, ESAK (mGy) and Incident Air Kerma, IAK (mGy) for general radiography as well as $CTDI_{vol}$ (mGy) and DLP (mGy.cm) for CT.

For general radiography, IDRL covers several of anatomy-based procedures, namely abdomen AP, ankle joint AP, antebrachial AP, BNO (*Blass Nier Oversich*) AP, chest (AP and PA), cervical (AP and lateral), femur AP, genu (AP and lateral), lumbar spine (AP and lateral), manus AP, pedis AP, pelvis AP, shoulder, skull (AP and lateral), GR-Cruris/Tibia Fibula, wrist joint AP, and waters. For CT, the established DRL covers the following anatomy-based procedures; abdomen (contrast and non-contrast), abdomino-pelvis (contrast and non-contrast), cardiac studies (contrast), chest (contrast and non-contrast), head (contrast and non-contrast), neck (contrast), and urology (non-contrast).

The data used to establish IDRL is obtained from data reported online by hospitals or clinics to the integrated national system for patient dose data registry (Si-INTAN, https://idrl.bapeten.go.id). The data collection process has been ongoing since 2015 until present, that includes data on examination of patients with CT (available since 2015), interventional radiology (available since 2016), diagnostic nuclear medicine (available since 2017), general radiography (available since 2017), dental radiography (available since 2018), and mammography modalities (available since 2018).

From 2015 to 2020, the availability of adequate data to determine IDRL is data for examining patients using CT and general radiography. In 2021, the IDRL value for CT and general radiography has been established. In 2022, it is planned to determine IDRL values for interventional radiology and diagnostic nuclear medicine, and in 2023 for dental radiography and mammography.

The dosimetric calculation system used in Si-INTAN to determine the patient dose value is the radiation output data for each modality obtained from compliance test results. The unique radiation output value for each x-ray modality will be used to calculate the patient dose, namely IAK, ESAK, and DAP. While for CT modalities, using the $CTDI_{vol}$ and DLP dose indicators on the monitor console that have been validated through compliance test, which has an error of less

than 20%. Another aspect considered for determining IDRL of adult patients is using the Indonesian adult patient weight standard (60 ± 10 kg), which was obtained from the analysis of body weight data from Si-INTAN for the period 2015 - 2018 and updated in 2019 - 2020. Dose indicators used in determining the IDRL value in Indonesia is enlisted in Table 1. (From the website www.idrl.bapeten.go.id):

Table 1 Dose quantities used in determining Indonesian Diagnostic Reference Level (IDRL)

No.	Modality	Dose Quantity	Derivative Indicators
1.	General radiography	ESD (mGy) or DAP or KAP (mGy.m^2)	Effective dose (mSv)
2.	Mammography	IAK (mGy)	Mean Glandular Dose (mGy)
3.	Fluoroscopy/ image-guided interventional	DAP or KAP (mGy.m^2) or Peak Skin Dose (mGy) or Air Kerma Rate (mGy/s)	Effective dose (mSv)
4.	CT	$CTDI_{vol}$ (mGy) or DLP (mGy.cm)	Effective dose (mSv)
5.	Intraoral dental	ESD (mGy)	Effective dose (mSv)
6.	Panoramic dental	DAP or KAP (mGy.m^2)	Effective dose (mSv)
7.	Nuclear medicine	Administered Activity (MBq)	Effective dose (mSv)

The establishment of IDRL is a mandate from the Government Regulation Number 33 Year 2007 on the Safety of Ionizing Radiation and the Security of Radioactive Sources. Furthermore, in 2020, BAPETEN Regulation Number 4 of 2020 was issued concerning Radiation Safety in the Use of X-Ray Equipment in Diagnostic and Interventional Radiology, which requires the licensee to submit patient dose records online through the national information system for patient radiation dose. According to the regulations, recording and reporting patient radiation dose data to Si-INTAN is one of the tasks of medical physicists. Therefore, it is necessary to encourage the participation of medical physicists to be active in the mechanism for collecting, reporting, and analyzing patient dose data to Si-INTAN, so that the national program in determining and implementing IDRL is easily realized.

Lao PDR

Ms.Viphaphone Inphavong proposed the survey for the local diagnostic reference levels, LDRLs, at Mittaphab Hospital, Vientiane Lao PDR. The survey includes CT and General X-ray examinations. The patient selection was at 18-70 years old, the weight range was 45-75 kg. The CT examination includes Brain with contrast (2-phase) /without contrast, Chest with contrast (2-phase) /without contrast, Abdomen with contrast (3-4 -phase), CT TAP (thorax-abdomen- pelvis) (3-phase). Dose quantity is $CTDI_{vol}$ (mGy), total DLP (mGy.cm). The median and the third quartile of the patient dose quantity were calculated from 289 cases. The General X-ray examination includes Sinus/Skull, Chest PA, Abdomen AP, Pelvis AP, Lumbar spine AP/ LAT. The dose quantity is KAP (μGy.m^2), the median and the third quartile of the patient dose quantity were calculated from 178 cases.

Malaysia

Ms.Nurmazaina Md.Ariffin proposed the survey of CT, General X-rays, Fluoroscopy, and Interventional Radiology. 277 CT systems, 2,457 X-ray systems, 718 Fluoroscopy systems, and 170 IR systems are in the country inventory. The range of patient weight is 40-80 kg. The dose quantity for CT is $CTDI_{vol}$ (mGy) and DLP (mGy.cm), the examination consists of Abdomen, Brain, Cardiac, Chest, Pelvis, Spine/ Musculo- Skeletal, Thorax. The dose quantity of general X-rays is Air KERMA (mGy) and KAP (mGy.cm^2). The examination consists of Abdomen (KUB) AP, Cervical Spine AP/LAT, Chest PA/ LAT, Extremity upper/lower, Lumbo-sacral AP/LAT, Pelvis AP, Skull AP/PA, Skull LAT, Thoracic spine AP /LAT. The dose quantity of fluoroscopy is the Cumulated Air KERMA (mGy), KAP (Gy.cm^2). The examination consists of ERCP, GI upper, GI Lower. The IR examination consists of cerebral, ESWL and vascular.

Myanmar

Ms. Thinn Thinn Myint proposed the survey of Nuclear Medicine, planar imaging, SPECT /CT, and PET/CT. There are six nuclear medicine centers, four are public centers and two are private centers. There are six SPECT/CT, two PET/CT systems, and one cyclotron. Nuclear Medicine examinations are planar and tomographic studies. Planar imaging is Thyroid, Parathyroid, Renal DTPA, Renal DMSA, Liver/Spleen, Hepatobiliary, and Bone. SPECT/CT imaging is Myocardial Perfusion (Rest and Stress), Parathyroid and Bone. PET/CT is an oncology study of F-18 FDG. At the public hospital, the patient injected activity has been recorded the fixed activity for individual examinations that based on the range of activity, i.e. Myocardial Perfusion (Rest and Stress), Tc-99m MIBI 8-10 mCi. PET/CT examination 18F-FDG 0.12 mCi/kg. At the private hospital, all injected activities had been recorded. Six nuclear medicine systems were proposed for NDRLs, those are Cardiac, Endocrine, Genitourinary, Haematological, Skeleton, Oncology systems. The survey will be started in 2022 when the delivery of the radiopharmaceuticals is available. Two centers will be in the survey. Those are Yangon General Hospital (Public) and Pinlon Hospital (Private).

The Philippines

The Food and Drug Administration and the Philippine Nuclear Research Institute, the two radiation regulatory bodies for ionizing radiation, are leading the establishment of National Diagnostic Reference Levels for the Philippines. The national policy for the establishment of DRLs is currently being finalized. Implementation and data gathering shall follow once the policy has been formally issued. The goal is to establish the DRL for CT, Fluoroscopy, IR, General Radiology, Mammography, and General Nuclear Medicine. There are 354 facilities with CT systems; 30 CT systems will be included in the survey from 30 CT facilities. The range of body weight is from 50 to 70 kg. The dose quantities to be determined for CT are the $CTDI_{vol}$ (mGy) and the DLP (mGy.cm). The CT examination based on clinical indication will consist of Head (Acute Stroke), Chest (Lung Cancer), Chest (High Res), Abdomen (Liver Metastasis), Abdomen and Pelvis (Abscess), Chest Abdomen Pelvis (cancer), CT-Stonogram, and Head (Trauma). The number of general radiology facilities is 2593 of which 30 are to be included in the survey. The dose quantity to be measured is the Air KERMA (mGy). General X-ray examinations to be covered are Chest/Thorax (PA), Cervical Spine, Thoracic Spine, Lumbar Spine/LSJ, Abdomen, Pelvis and Hip, Skull (AP/PA, LAT), Pediatric - Chest (Lung), Pelvis, Abdomen, Skull (AP/PA, LAT), and Babygram. There are 109 facilities doing Fluoroscopy examinations of which 30 are to be included in the survey. The fluoroscopy examinations to be covered are Barium Meal, Barium Enema, Barium Follow Through, Intravenous Urography, and ERCP. There are 49 facilities doing IR of which 20 are to be included in the survey. The IR examinations to be covered are Coronary Angiography, Percutaneous Coronary Intervention, CA and PCI. There are 137 mammography centers, and 30 centers are to be included in the survey. The range of patient weight is 45- 65 kg. The dose quantity to be determined is the Mean Glandular Dose (MGD, mGy), and the examination to be covered is the screening mammogram.

Singapore

Mr.Kwok Yew Mun and Mr.Somanasen S proposed the survey of CT, IR and General Nuclear Medicine. The number of CT systems is 120, the number of department with CT is 20. The number of CT system in the survey is 8. The range of patient weight is 55-75 kg. The CT examination consists of Brain, Abdomen and Pelvis, Chest HRCT, Liver, multi-phase, KUB, CTA Circle of Willis. The number of IR X-ray systems is 78 from 10 departments; the data from 5 systems at a department is in the survey. The dose quantity is Cumulated Air KERMA (Gy) and KAP ($Gy.cm^2$). The IR examination consists of AVF Angioplasty, AVG Thrombolysis/Thrombectomy, CVC Tunnelled (Dialysis/pheresis), ERCP (Fluoroscopy only), Percutaneous Change of Catheter, Peripherally Inserted Central Catheter. There are 10 nuclear medicine departments and one department is in the survey. The dose quantity is the administered activity (MBq, mCi) and administered activity/ body weight (MBq/kg).

Thailand

Prior to the survey of the patient dose from medical imaging, the Section of Radiation and Medical Devices, arranges the training program to radiology and related staffs from Department of Medical Science on dosimetry based on IAEA TRS 457 and the concept of optimization of protection in the medical exposure of patients for diagnostic and interventional procedures. Radiology equipment and dosimeters in Thailand are calibrated annually and certified by Department of Medical Sciences, Ministry of Public Health. Under Certification, the radiology equipment is eligible for clinical service.

The survey was started in 2017 on General radiology and Dental radiography, 2018 on Computed Tomography, 2019 on Screen Film and Digital Mammography, 2020 on Fluoroscopy, Interventional Radiology and Cardiology. General Nuclear Medicine, SPECT/CT and PET/CT are surveyed by Nuclear Medicine Society of Thailand. In 2021, Department of Medical Science announced the establishment of Thailand DRLs 2021. The milestone in 2022 will be the survey on panoramic radiography and Cephalometric radiography, 2023 – Cone Beam CT for dental radiograph, 2024- Digital Mammography and Digital Breast Tomosynthesis.

The survey of interventional cardiology based on clinical indication, is performed by the Cardiovascular Intervention Association of Thailand. Number of intervention cardiology PCI cases was 22,737from 76 X-ray systems in 38 departments, 18- month survey with registry. The patient BMI, mean $\pm$ SD is recorded. Dose quantity is Cumulated Air KERMA (mGy) and KAP ($Gy.cm^2$). The examinations consist of Coronary Angiography (CAG), Percutaneous Coronary Intervention (PCI), Permanent Pacemaker (PPM), Chronic Total Occlusion (CTO) and non CTO.

There were 722 CT systems in Thailand in 2018 and 135 CT systems were in the survey. The size indicators were weight (kg), height (cm) and body thickness (cm). The range of body weight is 45-75 kg. The CT examinations were Brain without contrast media, Brain with contrast media, Chest without contrast media, Chest with contrast media, Whole abdomen without contrast media, Whole abdomen with contrast media.

546 from 8225 dental radiography systems were in the survey from 15 dental departments. Dose quantities were Incident Air KERMA, Entrance Surface Dose, Air KERMA at Isocenter (mGy), KAP ($mGy.cm^2$). The examinations of

Maxillary/Mandibular were Incisor, Canine/premolar, Molar.

250 from 22,275 General X-ray systems were surveyed on Chest PA, Lumbar spine AP, Lumbar spine LAT, Pelvis AP, Abdomen AP, Skull AP/PA, Skull LAT examinations.

1846 Fluoroscopy systems were in the survey of Barium swallow, Upper gastrointestinal fluoroscopy with contrast, Long GI, Barium enema and HSG examinations. 135 from 251 IR equipment were surveyed on TACE 2D, TACE 3D, PTBD, GI Bleeding, Diagnostic cerebral, Embolization of intracranial aneurysm, Embolization of brain AVM, Embolization of intracranial Dural AVF, Embolization of head & neck tumor.

166 from 456 mammographic systems were surveyed on screening and diagnostic mammogram on CC, MLO views of both breasts. Dose indicators were MGD and ESAK (mGy).

There are 32 nuclear medicine centers, 25 centers were in the survey of general nuclear medicine, Planar Imaging, SPECT/CT and PET/CT imaging. Planar imaging examination were Cardiology, Endocrine, Genitourinary, Gastrointestinal, Infection, Lymphatic, Oncology and Skeletal Imaging. Dose quantity was injected activity, mCi and mCi/kg. 34 SPECT/CT and 13 PET/CT were in the survey of Myocardial Perfusion, Cardiovascular, Lymphatic (Breast Ca), Neurological, Pulmonary and skeletal examination. PET Imaging is on Neurology, Oncology and Whole Body scan of F-18 PSMA, F-18 FDG, Ga-68 DOTA-TATE, Ga-68 PSMA. CT DRLs were estimated and reported in $CTDI_{vol}$ (mGy) and DLP (mGy.cm) on the hybrid systems.

Table 2 The NDRLs Thailand on 10 examinations

General Radiography	ESD mGy	Intra Oral Radiography	K_i mGy
Chest PA	0.3	Maxillary Incisor	2.3
Abdomen AP	3.8	Canine/premolar	3.1
Pelvis AP	3.1	Molar	4.0
LS AP	3.8	Mandibular Incisor	1.9
LS Lat	9.8	Canine/premolar	2.4
Skull AP/PA	2.6	Molar	3.1
Skull Lat	2.1		

Computed Tomography	$CTDI_{vol}$ mGy	DLP mGy cm
Brain without CM	1028	62
Brain with CM	935	52
Chest without CM	417	18
Chest with CM	665	18
Whole abdomen without CM	717	18
Whole abdomen with CM	717	20

Mammography	D_G mGy
MGD at 45 mm PMMA	2.50
2D MGD	2.04
ESAK	9.70

Interventional Body Radiology	KAP Gy.cm^2
TACE CBCT)3D(	226
TACE)2D(	141
PTBD	13
GI bleeding	151
Interventional Neuroradiology	KAP Gy.cm^2
Cerebral angiogram	108
Embolization of intracranial aneurysm	209
Embolization of brain AVM	187
Embolization of brain AVF	261
Embolization of Head & Neck tumor	230
Embolization of spinal AVM or tumor	210
Interventional Cardiology	**KAP Gy.cm^2**
Angiography coronary arteries CAG	28
Percutaneous coronary intervention PCI	99
Permanent pacemaker PPM	8

Nuclear Medicine System (SPECT)	Radiopharm	Activity mCi	Activity/BW mCi.kg^{-1}
Skeletal, Bone Marrow	Tc-99m MDP	21	0.44
Cardiovascular	Tc-99m MIBI (rest)	10	0.35
	Tc-99m MIBI (stress)	26	0.51
	Tc-99m PYP	20	0.30
	Tc-99m RBC (MUGA)	21	0.63
	Tl-201	3	0.05
Oncology	Ga-67 Citrate	5	0.09
	I-131	3	0.11
	I-131 MIBG	1	0.07
	Tc-99m MAA	5	0.11
	Tc-99m MIBI	20	0.40
	Tc-99m Octreotide	20	0.25

SPECT/CT Part of CT	Radiopharm	$CTDI_{VOL}$ mGy	DLP (mGy.cm)
Cardiovascular	Tc-99m RBC (MUGA)	2	200
Myocardial Perfusion	Tc-99mMIBI	2	200
Neurology	Tc-99m ECD	60	100
Oncology	Tc-99m Octreotide	12	400

PET System	Radiopharm	Activity, mCi	Activity/BW mCi.kg^{-1}
Neurology	F-18 DOPA	10	0.14
	F-18 FBB (amyloid)	11	0.19
	F-18 FDG	7	0.15
Oncology	F-18 FDG	9	0.21
	F-18 PSMA	6	0.12
	Ga-68 DOTA-TATE	5	0.07
	Ga-68 PSMA	5	0.09

PET/CT Part of CT	Radiopharm	Activity, mCi	Activity/BW mCi.kg^{-1}
Neurology	F-18 FDG	36	688
Oncology	F-18 FDG	11	500
	F-18 PSMA	13	500

Vietnam

There are more than 6000 X-ray facilities for General X-rays, Fluoroscopy, Dental, Mammography Intervention Radiology, and CT examinations. About 30 nuclear medicine centers are serviced in Vietnam.

At Cho Ray hospital, Ho Chi Minh City, further from general x-rays, there are 8 CT scanners, 2 PET/CT, 2 SPECT and SPECT/CT.

In 2022, the survey of common CT and Nuclear Medicine examinations will be started for local diagnostic reference level, LDRLs, such as CT Head, CT Abdomen, three phase liver, and etc. The dose quantity for CT protocol is DLP (mGy.cm). For Nuclear Medicine, the dose quantity is the injected activity in mCi for planar and tomographic imaging. PET/CT imaging, the dose quantity is administered activity per patient body weight at 0.12 mCi/kg of ^{18}F-FDG. For each examination, the data collection is at least 30 patients per examination.

Conclusion

Even though the European Union started the RDRLs in 1997, the regional diagnostic reference levels in ASEAN had just been started in 2021 with large variation in the survey. The major problems are from the lack of knowledge and experience of the personnel in medical imaging especially the medical physicist. The lack of nuclear medicine and the interventional services in Lao PDR resulting in the Lao patients cross Mae Kong River to obtain the service in Thailand. Such the culture on the neighbor supportive has been extended to the education, clinical training to be sustainable in the future. The national and regional diagnostic reference levels are on the voluntary basis, the encouragement and the recognition on the optimization of radiation protection in medical imaging are emphasized by the consultants. There may be a lot of pitfalls in the first RDRLs of ASEAN – ADRLs, but the lesson learned can improve the next survey of ADRLs in the standards of the ASEAN surveys especially the inclusion of the clinical indication in the survey.

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REPORT ON THE VIRTUAL MEETING OF 19th SEACOMP, 14th ACOMP and 13th TMPS

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Abstract— **Thai Medical Physicist Society (TMPS) hosted the virtual meeting of the 19th South-East Asian Congress of Medical Physics (SEACOMP), the 14th ASEAN College of Medical Physics (ACOMP) and the 13thAnnual Meeting of TMPS under the theme "Medical Physics – Facing the Future Together". The congress was held on 21st – 23nd October 2021. There were 231 online delegates from 16 countries participating the congress. The 1st SEAFOMP-Anchali Krisanachinda Medical Physics Quiz" was organized at this congress.**

Keywords— **SEACOMP, TMPS, ACOMP, Virtual Congress.**

I. Introduction

At the 18th SEACOMP Council Meeting in 2020 at Phuket, Thailand, Thai Medical Physicist Society (TMPS) had been asked to host the 19th South-East Asian Congress of Medical Physics (SEACOMP) in Phuket Island, Thailand again to extend the opportunity to the participants to visit Phuket Island. However, the rapid COVID-19 pandemic in ASEAN countries has a huge impact on a face-to-face scientific meeting. The congress was adapted to a virtual meeting and held at King Chulalongkorn Memorial Hospital, Bangkok, Thailand.

II. Details of the conference

Only the local committee members, medical physics students, and the IT persons were onsite to organize the online congress under the WebEx platform including three sponsor companies as presented in Fig. 1 and Fig 2. The 19th SEACOMP, the 14th ACOMP and the annual meeting of 13th TMPS had been organized under the theme **"Medical Physics – Facing the Future Together"**. The aim of this conference was to strengthen medical physics in healthcare and to explore the latest developments of medical physics in research, education, professional development, and industry.

There were 231 delegates from 16 countries participated in the conference virtually as in Table 1. Most of the participants were Thai but it also attracted delegates and invited speakers from other continents. This congress has 19 invited lectures, 10 mini-symposiums, 2 ACOMP lecture-workshops, and 8 commercial video presentations. The field of the congress are Medical Imaging, Radiation Therapy, AI, Education, and Clinical Training of Medical Physics. Moreover, 43 proffered papers were submitted for the oral presentation of 12 papers in radiation oncology, 17 papers in diagnostic radiology, 4 papers in nuclear medicine, and 10 papers in artificial intelligence presentations. All invited lectures and oral presentation abstracts including 16 full papers were published in the congress proceedings. The congress was well supported by many organizations and societies, such as the International Atomic Energy Agency (IAEA), American Association of Physicists in Medicine (AAPM), Japanese Society of Medical Physics (JSMP), Japanese Society of Radiological Technology (JSRT), and several universities as well as 18 healthcare industrial partners.

Fig. 1 The local organization team and medical physics students from Thailand

Fig. 2 The local organizer visited three sponsor companies

The opening ceremony was held on 21st October 2021. Dr. Anchali Krisanachinda, the Congress President gave the welcome speech and Dr. Freddy Haryanto, SEAFOMP President proceeded to the Opening Remarks. After that, the 16th John Cameron Memorial Lecture was started by Prof. Dr. Peter Homolka from the Medical University of Vienna, Austria, on the topic "3D printing in medical imaging: opportunities and potential pitfalls for the medical physicist" (Fig. 3). His lecture covered the application of 3D printing technology to medical imaging such as the phantoms for quality control, commissioning, and dosimetry or the higher-level phantoms of the pseudo-anthropomorphic phantom.

Table 1 The countries and number of participants.

Country of Participants	Number
Thailand	142
Indonesia	19
Malaysia	17
The Philippines	14
Japan	9
Singapore	7
USA	7
Austria	4
Australia	3
Cambodia	3
Canada	1
Lao PDR	1
Myanmar	1
Vietnam	1
Taiwan	1
Yemen	1
Total	231

Fig. 3 Prof Dr. Peter Homolka delivered the John Cameron Memorial Lecture on "3D printing in medical imaging: opportunities and potential pitfalls for the medical physicist"

The virtual meeting was organized in three tracks of physics in diagnostic radiology, radiation oncology, and nuclear medicine in parallel. During the coffee and lunch breaks, the pre-recorded commercial video presentations were displayed. The SEAFOMP EXCOM meeting was organized at lunchtime on October 23rd, 2021. SEAFOMP committee made the recommendations for SEAFOMP activities last year, the SEAFOMP Treasurer report 2021 was approved.

The 14th ASEAN College of Medical Physics (ACOMP) in AI and targeted radionuclide radiotherapy was traditionally organized in this congress. The AI course was announced in the poster in Fig. 4, consisting of 7 lecture-workshop topics organized by Professor Kwan Hoong Ng and his team.

Fig. 4 Workshop poster of 14th ACOMP course

Prof. Kwan Hoong Ng, University of Malaya, proposed the SEAFOMP Quiz at the congress to strengthen the young and active medical physicists, to encourage the collaboration, and the bonding among SEAFOMP members. "SEAFOMP-Anchali Krisanachinda Medical Physics Quiz" is an official program on this congress. The committee members were Prof. Kwan Hoong Ng and Dr.Yeong Chai Hong from Malaysia, Dr.Nur Rahmah Hidayati from Indonesia, and Dr.Taweap Sanghangthum from Thailand served as quiz masters. The quiz was held on the last day before the closing ceremony through the Kahoot platform. Two junior medical physicists from SEAFOMP members, Cambodia, Indonesia, Laos, Malaysia, Myanmar, the Philippines, Singapore, and Thailand participated the quiz. Twenty from thirty questions were on medical physics topics and ten questions were on SEAFOMP-related topics. Fig. 5 shows the first, second, and third winners named Mr. Jacobs Mata from the Philippines, Miss Mananchaya Vimolnoch from Thailand, and Miss Sararas Intarak from Thailand, respectively.

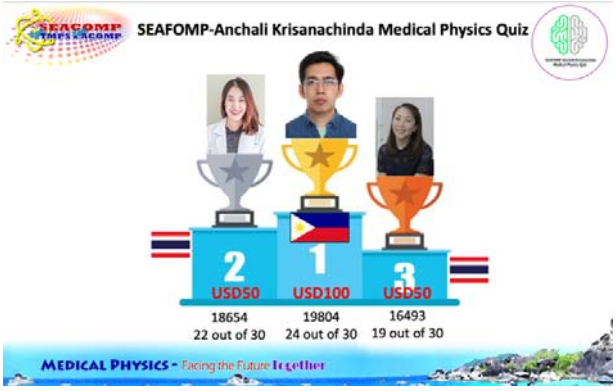

Fig. 5 Three winners for SEACOMP-Anchali Krisanachinda Medical Physics Quiz

After the quiz, Dr. James Lee, the Co-Chair of the local scientific committee of the IUPESM- World Congress on Medical Physics & Biomedical Engineering (IUPESM WC2022) and the 20th SEACOMP had announced the mentioned congress which will be held hybrid on 12-17 June 2022 in Singapore.

Dr. Anchali Krisanachinda gave the closing remarks including the announcement on the SEAFOMP oral presentation awards. The best presenters in radiotherapy was Miss Dainna Recel Pamisa from the Philippines, in diagnostic radiology were Miss Riska Amilia from Indonesia and Mr. Abd Halim Mohd Fadhullah from Malaysia, in AI was Miss Sa-angtip Netprasert from Thailand. After that, the president announced the 4th SEAFOMP young leader awards (YLA) in recognition of the outstanding achievement of young medical physicists in the region. There were 4 medical physicists, Ms. Indah Lestariningsih from Indonesia, Dr. Aik Hao Ng from Malaysia, Ms. Julie P. Cruz from the Philippines, and Dr. Suphalak Khachonkham from Thailand (Fig. 6). Finally, the congress was adjourned with a group photo, online and onsite together (Fig. 7).

(a) (b) (c) (d)

Fig. 6 The fourth young leader awardee, 4th YLA, (a) Ms. Indah Lestariningsih from Indonesia, (b) Dr. Aik Hao Ng from Malaysia, (c) Ms. Julie P. Cruz from the Philippines, and (d) Dr. Suphalak Khachonkham from Thailand.

III. Conclusion

The congress has successfully finished with the support from many organizations. We acknowledge with thanks all the organizations, SEACOMP & ACOMP 2021 committee members, the distinguished invited speakers, as well as all colleagues who contributed and participated at the SEACOMP 2021 virtually. Last but not least, the appreciation for the continuous support from the local and regional vendors. The truly international friendly spirit at the SEACOMP 2021 was one of the main pivots of the success of SEACOMP 2021.

Fig. 7 The group photo at the closing ceremony

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HOW-TO

STRATEGIC PLANNING: CASE STUDY FOR A DIAGNOSTIC RADIOLOGY CONSTANCY TESTING PROGRAMME IN A MAJOR HOSPITAL IN MALTA

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Abstract— **Medical Physics is a young profession in Malta and the present constancy testing programme at the hospital concerned, younger still. It was therefore felt necessary that a strategic evaluation of the current state of the programme be carried out. The objectives were: (a) to develop a vision statement for the constancy testing programme in diagnostic radiology which complies with the role of the hospital within Maltese society and healthcare system (b) to carry out a SWOT thematic analysis to evaluate the current programme, and (c) to provide recommendations for improvement in terms of a list of strategic objectives. Using a qualitative research approach, data were collected by means of semi-structured interviews with the Medical Physics professionals involved in the programme, direct observations and document analysis. The current programme has several strengths but also several weaknesses, mainly derived from the profession being so young in Malta. Fortunately, several opportunities for programme improvement are available, however, some threats do exist. By keeping patient service in mind and taking strategic management approaches, continuous quality improvement of the constancy testing programme can be assured.**

Keywords—**Constancy testing programme, evaluation, medical imaging devices, quality improvement, SWOT.**

All statements made regarding the operations at the hospital concerned or the findings of the study reflect the opinions of those interviewed.

I. Introduction

An effective constancy testing programme for medical imaging devices is crucial for ensuring high-quality images, high diagnostic accuracy and optimising patient doses. Such programmes should be based on general quality management standards [1-3]. A strengths, weaknesses, opportunities, threats (SWOT) analysis is vital for programme quality improvement. SWOT-based strategic planning is used widely in various fields, and healthcare is no exception. Examples include: role development, inter-professional healthcare education at academic medical centres and establishing individual medical strategies for patients with aortic disease, to name a few [4-6].

The Medical Physics (MP) profession is young in Malta, and the present constancy testing programme at the hospital concerned, younger still. So far, no formal, systematic strategic evaluation of the present state of the programme including recommendations for future improvement has been carried out. At present, five MP professionals (MPPs) in the specialty area of Diagnostic and Interventional Radiology (D&IR) manage the constancy testing programme for the modalities in this area. The modalities are: general projection radiography, CT, fluoroscopy, angiography, dental imaging, mammography, bone densitometry, MRI and ultrasound. By working with radiologists, radiographers and engineers, MPPs ensure acceptable performance of such devices.

The objectives of the study were to develop a vision statement for the constancy testing programme for the hospital concerned which is consonant with the role of the hospital within Maltese society and healthcare system, to evaluate the current programme with respect to the developed vision via SWOT analysis and to make recommendations for programme improvement in terms of a list of strategic objectives. The research study provided a golden opportunity for the present MPPs to reflect on their current practices.

II. Materials and methods

Policy statements published by MP organisations, relevant quality standards, European legislation, and other relevant documentation published by standard setting organisations, MP professional organisations, the IAEA and the European Commission (EC) were consulted in order to develop an appropriate vision statement for the constancy testing programme. An inventory of SWOT themes was then developed by means of Semi-Structured Interviews (SSI) with the MPPs involved, observations at the hospital concerned and document analysis of associated documents. All current MPPs participated in the SSIs. The interviews were audio-recorded following their consent; every audio file was transcribed and then the recordings destroyed to ensure data protection. The tool used was a SSI data sheet which was developed in such a way as to encourage the emergence of SWOT themes. For the on-site observations, an extended time period was spent in the hospital to gain in-depth understanding of the tests performed, the working environment and the people involved . Reflective notes were recorded [7].

Data analysis was performed by developing a 'thematic template' to categorise SWOT themes from the data so that useful information may emerge. Template analysis was carried out using NVivo software. NVivo helps code the data in an efficient manner while identifying and organising

suitable themes effectively [8]. For document analysis, data were imported in NVivo and file classifications were used to classify documents accordingly. For the interviews, the data were coded and analysed in a way that results could not be traced back to the individual participants.

III. Results and discussion

This section provides the proposed vision statement for the constancy testing programme and a discussion of the results of the SWOT thematic analysis derived from the interviews, observations and document analysis. External opportunities and threats of the programme were further subcategorised as political, economic, social or technological-scientific (PEST). Every SWOT theme was rated from +1 (low) to +10 (high) for S/O themes and −1 (remote) to −10 (very severe) for W/T themes based on their potential impact on the achievement of the vision. Strategies for achieving the desired vision are also suggested.

Vision statement

The following vision statement expresses the desired future state of the constancy testing programme based on international standards and aspirations of the D&IR team at the hospital concerned:

"Our constancy testing programme will be recognised by all stakeholders as an optimal (effective and time/cost efficient) and comprehensive (i.e., all available modalities tested) constancy testing programme structured in line with internationally accepted guidelines and aimed at ensuring patient safety through the use of the most updated constancy testing protocols and managed by fully-qualified medical physics professionals."

STRENGTHS of the programme

1. Large number of tests performed *(+8)*

The MPPs perform a lot of tests when compared to most countries and hence find a sense of accomplishment, pride, and dedication. These are feelings based on their drive to 'prove' themselves to hospital management and other healthcare professionals (HCP) and make these stakeholders aware of their importance in the hospital and of their scholarly achievements. This drives the participants to improve the quality of the programme. A MPP said, "when we attend international courses, I realise we're quite advanced compared to other countries".

2. Appropriate scheduling of tests *(+2)*

With appropriate test scheduling, risks of costly mistakes are reduced and overall outcomes improved. Tests and tasks that need to be prioritised are easily identified. Given the high number of medical imaging devices in the hospital, the team manages to perform all required tests, though owing to time constraints, perhaps not all to the required level.

3. High quality protocols used *(+7)*

The MPPs pride themselves in using the most up-to-date constancy testing protocol standards that provide appropriate action and tolerance limits, which in turn translate to suitably tight remedial and suspension levels. A participant said, "We feel that the manufacturer's tolerances are sometimes too wide for us to accept during constancy testing because the risk to the patient would be high". On the other hand, another MPP said that the procedure of a standard being bought by the hospital procurement unit sometimes simply took too long, "we need constancy testing standards for the new technology, which we don't always have. The procurement process is sometimes so slow, it's very frustrating".

4. Testing in small teams *(+5)*

The MP team perform constancy tests in groups of two/three. This helps them optimise the time, do tests faster and ensure that the device is not taken out of service for long. It has also been noted that being in teams helps detect certain errors that a single MPP may perhaps not notice.

5. Enough time to perform the testing *(+6)*

The time taken to perform the constancy tests on the devices may take hours to complete. When MPPs request a number of hours allocated for the tests, often (though not always), the radiology team cooperate and "give up the machine". A MPP said, "For the most part if we tell them four hours, they do give us four hours. At most they allocate another time on another day for us, but when we request a time, they give it to us. However, there are exceptions."

6. MPPs feel a general sense of recognition *(+2)*

MPPs feel a sense pride when receiving recognition from other HCP. This makes them more motivated when conducting the tests, as a MPP said, "a radiologist told us, "If it weren't for you, I would have thought there was an issue with the patient, and we would have sent him for further tests". That made us feel that our tests are important and that we're making an impact." Recognition of efforts increases the quality of the constancy testing programme as it translates into greater enthusiasm and motivation for testing.

7. MPPs have high analytical and problem-solving skills *(+10)*

MPPs use their renowned problem-solving and troubleshooting skills in reporting and tackling problems that arise with the recognition of less-than-optimal performance of devices. A MPP said, "we don't run away from problems but face them with open arms as we embrace challenges… that's how we were taught". With

the different background they have from radiologists and radiographers, such as looking at an image from a different perspective which is more quantitative, they feel they are important and of great value as part of the multidisciplinary team.

8. Excellent MP training *(+9)*

The participants had common training according to the official recommended IAEA training programme as required by EU guidelines, at a single centre in the UK [9-11]. Therefore, the MPPs feel they have a shared knowledge between them and know where the team's skills and competences stand. A MPP highlighted that they could speak openly in the team and can perform the tests faster together as a result of this commonality.

9. Determination, competitivity and drive for improvement *(+4)*

The MPPs feel they must strive for higher effectiveness and efficiency so that other HCP may appreciate their work and be aware of the importance of the constancy testing. They are determined to make their presence felt and feel that this must remain so and founded on a cycle of continuous improvement.

10. High knowledge, skills and competences with respect to medical imaging devices *(+10)*

MPPs have high expertise in the safe, effective and scientific use of medical imaging devices and deep understanding of their performance which adds value to clinical problem solving and the technology assessment process. A MPP said, "I feel that the expertise that we have in understanding machines gives us the advantage that we are of value to the diagnostic team". The educational programmes have included units about quality management, professional issues and metrology amongst others, which are directly relevant to constancy testing of medical imaging devices. This gives MPPs the capability of suggesting and applying new technologies. D&IR MPPs have a leading role in the establishment and evaluation of criteria that determine acceptable performance.

11. High competences in radiation patient safety *(+8)*

Given their E&T regarding safety and protection from physical agents, MPPs work with the mindset of patient safety which helps them be extremely careful when conducting the tests, since they understand the subsequent risks involved.

12. High competences in undertaking clinical research *(+5)*

MPPs learned to undertake research frequently in their educational studies since as scientists, they were taught not to 'trust their instinct completely' but provide proof for their claims. Thus, MPPs are fit for carrying out clinical research. This is important, to keep up with the advancements of medical technology and update the constancy testing protocols. Since constancy testing is increasingly becoming quantitative, MPPs' quantitative approach is becoming more valued.

13. Ongoing update of testing protocols, equipment, and test objects *(+8)*

Investment in newer medical technologies leads to direct increases in patient health outcomes. By using newer equipment, MPPs find it easier to conduct tests with increased precision: "nowadays the devices have developed so much that high precision is easily achieved". MPPs keep their constancy testing documents up-to-date and ensure that the process of procurement of QC standards documents is ongoing.

WEAKNESSES of the programme

1. Internationally available standards not always keeping up with progress of medical technology *(−8)*

Most MPPs feel that the internationally used standards they follow to carry out constancy tests progress at a slower rate than the progress of the medical imaging devices at the hospital, so that certain tests which were applicable years ago are not applicable anymore. One MPP stated, "we might be testing something that today does not need any testing, but we cannot be too sure… so it's best to take note and see the context in which these standards had been originally written. On the other hand, there may be tests which are currently not being done but need to be introduced". MPPs also experienced the situation where some tests provided in internationally used standards cannot be carried out because of 'newer' features on the devices at the hospital. Sometimes there has been insistence by authorities on the use of international standards which are outdated, restricting the highly competent professional from introducing innovative techniques in the process.

2. Subjectivity of certain tests *(−6)*

MPPs feel that the element of subjectivity in some constancy tests is too high since some are based on qualitative visual scoring. This makes their comments in reporting not as convincing as when based on objective measures. They feel that this weakens the possibilites of constancy testing. Vague phrases such as "at least one pixel width should be resolvable" [12] and "count the number of grayscale steps" [13] do not provide solid foundations for reliable constancy testing. Automated analysis techniques may eliminate the variability in the results due to the variability in human observer performance [14].

3. Difficult access to devices to perform tests *(−6)*

The participants highlighted that it is sometimes not easy to schedule tests with radiologists which at times results in

difficult access to the devices. Even when managing to book time on a device, there is sometimes the need to set up the equipment for testing all over again because of a device that is needed urgently by radiologists. A MPP said, "we either continue after the exam or reschedule the device for testing". This results in a lack of consistency in the results, even for a particular machine. Another MPP said with disappointment, "You start doing the tests, then something crops up and you have to put everything on hold".

4. Lack of proper handover *(−3)*

There is a lack of clarity when it comes to handing over of annual constancy tests of the devices. A MPP said, "Today I did something myself, the next year another person does it and the year after someone else. There is perhaps not sufficient coordination in the handover sometimes – I do something today and then the job isn't mine anymore, the next person will take care of it." This lack of insufficient long-term scheduling, low feedback and communication leads to poor forecasts of future device trends, loss of time and lower quality for the constancy testing programme. Yet this weakness is not only present among MPPs, but it afflicts many hospital departments.

5. Lack of well-defined acceptance criteria for some tests *(−8)*

For some tests, there are no well-defined thresholds and limits - typically for non-ionising modalities, which may be due to the absence of specific legal requirements to conduct such tests [12]. For example the term 'significant' is often used too loosely - each user must select their own threshold above which they consider a nonuniformity significant [13]. A similar ambiguity exists for spatial resolution tests; "The measurement of spatial resolution (and threshold contrast) is subjective". This weakness defeats the much-needed objectivity in the constancy testing leading to loss of trust in the constancy testing programme as a whole. A MPP also said, "I know something is going on, but I cannot provide proof because there is nothing sufficiently convincing to back me up."

6. Low number of MPPs *(−10)*

The MPPs said there was a lot to be done but the understaffing issue limits the growth of the constancy testing programme and hence there is only so far they can go with programme improvement. The inadequate staffing levels result in the hospital not experiencing the full benefits of the constancy testing programme.

7. Low research activities *(−3)*

Research is limited to when MPPs find time. A MPP said, "time is precious here. We would like to increase research activities, because there is a lot we can expand and improve on." Low research limits expansion and progress of the constancy testing programme since 'outdated methods' are used on 'newer technology' because there is not much time available for them to improve the tests via research. Innovation is therefore often stunted.

8. No or low specialisation *(−5)*

The low number of MPPs available and the high number of devices for testing, limits MPPs from subspecialising into separate modalities. Some MPPs do not manage to understand the reasons behind a fault due to their lack of knowledge about the device and cannot master their understanding because there is no time for doing so. Subspecialisation of MPPs would make available exquisite expertise which would produce a quantum leap in the optimised constancy testing and use of the devices. Lack of subspecialisation may lead to poor programme quality as a participant stated, "Every week you're working on a different device and you start getting confused as to which tests are applicable to which modality." All the MPPs showed interest in being able to subspecialise as they feel that this has become a necessity given the rapid expansion in the quantity and sophistication of imaging device technology - "it's simply impossible to be an expert of all the devices which we test".

9. Insufficient collective decision making *(−4)*

Most participants highlighted that there is insufficient cohesion within the MP team. Being such a small team, this may directly limit improvement of the constancy testing programme since the quality of the team is a highly essential influencing factor. The issues seem to be stemming from past historical events arising before the taking over of the present lead physicist. Through management by objectives, self-management and personal commitment and active member participation, it is possible for a team to overcome such barriers to improved teamwork and communication [15, 16].

10. Low competences in management, leadership and general 'soft skills' *(−3)*

MPPs often find it difficult to bridge the gap from the physics/engineering environment to the economically and politically driven healthcare professions that they need to work with. MPPs are plucked out of the objective view of science into a more subjective, emotional and opinionated viewpoint in the hospital. Leadershp and soft skill training for MPPs is considered important for programme improvement [17].

11. Missing tests *(−6)*

Due to the absence of legal requirements, certain tests, particularly in non-ionising radiation are not done. This is an issue arising from the low awareness of the importance of the tests from the national Commission for the Protection from Ionising and Non-Ionising Radiation and the understaffing issue. However, the EC highlights the importance of tests on non-ionising radiation devices, since the link between non-ionising and ionising radiation

modalities is on the rise. The MPPs expressed the opinion that they would like to do the tests when they have enough manpower to do so as a MPP stated, "sometimes you need the strength of numbers to convince and we are still too few, but hopefully we'll get there."

12. Mixed levels of IT skills *(−3)*

Most MPPs highlighted that IT is important for the future of constancy testing to promote more objectivity in the test results. Knowledge of IT is a strong asset for MPPs since healthcare is becoming increasingly software based. But not all MPPs have these skills - this might limit progress.

13. Insufficient system checks *(−2)*

System checks, which include movement mechanisms, display options, image archive and networking capabilities to PACS, soft-copy and/or hard-copy locations seem to be often overlooked, as was evident from the observations. A MPP highlighted during a mechanical check for an imaging device, "I don't know whether this device's control panel is adjustable or not; it looks stuck, but I'm not sure".

14. No marketing among stakeholders *(−8)*

MPPs step back from publicising and advertising their services to stakeholders; there is an absence of a marketing strategy, which should not only be there, but should be ongoing [17]. Yet, all participants wish there was a higher level of awareness of their profession and role by stakeholders. A MPP stated, "when people ask me what I do for a living, I tell them engineering or applied science because they don't understand." This shows that the MPPs do not know how to describe their role to stakeholders who are not in the field. Other professions in the hospital barely know about the existence of 'medical physics', let alone constancy testing of medical imaging devices. This is a risk for the profession itself, as stakeholders will not know when to seek the advice or the services of the MPP.

OPPORTUNITIES for the programme

Political

1. General political pressure to achieve EU standards in all areas of the country *(+8)*

The general political climate in the country is pushing for the adoption of EU standards in all spheres of life. This drive helps provide an impetus towards more funding and human resources for constancy testing. MPPs stated that they would like to develop the programme in line with ISO 9001. The increase in testing frequency involves having updated SOPs for all modalities with correct versions and referencing. This is an opportunity for a quality jump in the programme as the ISO 9000 family of quality management system (QMS) standards has earned a global reputation as a basis for developing effective and efficient QMSs. ISO TG 176 and BS 70000 provide a step-by-step process of the implementation of a QMS that can be adopted by MPPs for the constancy testing programme [18-20].

Economic

2. Increase in human resources: Traineeship programmes *(+10)*

New trainees are essential assets for the growth of the profession, hence a huge opportunity for current MPPs to expand their team and increase recognition and awareness in the hospital. A new cohort of Masters students is graduating. This opens the opportunity for increased human resources and helping hands, thus increasing time available for work on innovative projects and for improvement of the constancy testing programme.

Social

3. A growing population leading to increased use of diagnostic medical imaging *(+7)*

Attributed to a growing population, there is a dramatic increase in the use and scope of diagnostic imaging which creates pressure for an increase in imaging devices and hence constancy testing. To cater for this increase, an effective programme is required that calls for an urgent need of characterising patterns of imaging use, re-defining "radiology services utilisation" and researching the role of constancy testing in such a milieu.

Technological-scientific

4. AI, test automation and objectivity *(+8)*

Today's world is moving towards AI, to reduce the time taken to perform repetitive tasks, to increase objectivity in the results while reducing human error. Automation has several benefits such as higher quality, improved worker safety, increased workflow rates and professional prestige. The physicists at Mayo clinic in the USA have developed software with algorithms based on digital image processing techniques, to automatically analyse test object images objectively. As an opportunity, a MPP suggested the use of AI to do automatic constancy checking, and the MP job would be performing QC on the AI rather than on the physical devices proper. Another MPP stated that knowledge in IT helps give a strong reputation to the profession in the clinical environment. Current MP postgraduate programmes in Malta include a 10 ECTS unit on Machine Learning (ML) and Pattern Recognition. Background in ML is considered a major strength in extending the role of the MPP.

5. Rapid prototyping ('3D printing') *(+1)*

Custom-made 3D printed test objects and phantoms made of tissue-mimicking material would increase constancy testing possibilities. 3D printing in-house would save on costs, promotes flexibility and drives innovation whilst reducing the carbon footprint.

THREATS to the programme

Political

1. Substitutability and over-commoditisation *(−7)*

MPPs fear substitutability and over-commoditisation with other HCP stepping in to do the MPPs' job regarding constancy testing of imaging devices. However, it is an undisputed fact that the competence of MPPs in such areas surpasses by far that of other HCP. It is up to MPPs to take a proactive stand against over-commoditisation which is not only a threat to the profession but also ultimately to patient safety [17].

Economic

2. Possible loss of team members *(−10)*

Since the MPPs trained overseas were bound by a finite contract as compensation for their training, nothing formally stops them from leaving the profession once the contract ends. Should this take place, the profession in Malta will become endangered since there are currently no trainees and hence no handover to a next generation. Thankfully, the highly successful B.Sc. (Hons.) in Physics, Medical Physics and Radiation Protection set up by the University of Malta is producing successive cohorts of potential human resources for the constancy testing programme.

3. Inadequate resources *(−9)*

Financial and material resources were also a running theme amongst the participants where a MPP said, "The CEO sees the radiologist as the responsible person for signing where money is involved" and hence this limits the acquisition of resources to what the radiologist perceives as necessary even though constancy testing is a responsibility of MPPs. This may limit the progress of the programme, however, there is a move towards setting up an independent MP department directly answerable to the CEO.

Social

4. Low feedback from the multidisciplinary team *(−7)*

The MPPs found that radiologists, radiographers and engineers rarely provided feedback about the functioning of the medical imaging devices after being constancy tested. In addition, owing to possible excessive professional pride or low awareness by other HCP, MPPs are not sought for advice before conditions become critical. Some MPPs experienced the situation of portable devices being transferred to other rooms by other HCP without being informed hence disrupting the constancy testing schedule of those devices.

5. Low awareness of the importance of constancy tests by radiology management *(−3)*

Most MPPs highlighted that the radiology team is not always sufficiently aware of the importance of constancy tests, or as a MPP said, "some are aware of their importance but won't give us time anyway". Another MPP said with frustration, "We studied all that in our educational courses, our hands are full of calluses from the difficulty of the subjects we studied, we've had all that training… you would at least want that they acknowledge what we do". The disappointment in the MPPs may result in risks to the programme since MPP satisfaction is an asset for improvement.

Technological-scientific

6. Insufficient knowledge of 'why' constancy tests fail *(−3)*

For physicists, not knowing why things develop the way they do can be a source of stress. A MPP said, "I'm testing the machine for image quality but I don't know how it does it. And then if it fails? I don't know where to start looking for the problem." The fact that there is currently an unclear definition of the essential parameters describing performance of medical imaging devices and how these may change with technological progress, makes it harder to categorise and understand failures . Not having the time to delve deeply into the devices which they test is a source of loss of self-confidence.

The way forward: Strategic Objectives

Based on this strategic SWOT evaluation exercise, the following strategic recommendations are suggested.

Further Strengthening of Internal Strengths

1. Attend international accredited courses on constancy testing to further develop existing competences and acquire missing ones.
2. Ensure further development of protocols by performing research to improve on or replace those internationally used protocols which are outdated.
3. Encourage team members to set up a table of key success factors of the constancy testing programme and rate them. The leader provides an average rating for the team and sets priorities accordingly.
4. Set up hands-on retraining programmes for all members of the D&IR team to ensure tests are performed as intended, and new/more updated protocols adopted.
5. Set up meetings to schedule discussions. Ensure objectives and goals are written and regularly updated on a board in a common area of the team's premises, as well as on a digital document to keep these visible to all members.

Reducing or Eliminating Internal Weaknesses

6. Develop a Gantt chart to assign time frames to action plans, and a Load chart to have a formal schedule of who needs to conduct which constancy tests and on which modalities. This ensures that the team keeps a continuous track of device trends.
7. Attend team building CPD to promote collaborative work.

8. Attend courses in quantitative approaches to quality management.
9. Develop and implement a profession awareness and marketing strategy.

Grasping External Opportunities

10. Involve oneself in reject analysis of patient images programmes. Where appropriate, link rejects to lack of constancy testing to increase awareness of the importance of the latter.
11. Attend MPP-designed IT courses so that constancy testing may be made more objective and limitations due to lack of IT skills eliminated.
12. Attend courses on CAD modelling and 3D printing to be able to produce custom test-objects.
13. Set up meetings with the radiology team to get to know more about patterns in patient imaging sessions. This helps strengthen scheduling of the programme.
14. Attend courses in leadership in medical physics.

Eliminating External Threats

15. Pro-actively combat counter-productive and unsafe substitutability and over-commoditisation.
16. Set up a feedback system between all team members and an inter-departmental feedback system with the multidisciplinary team. This enhances communication and clarification of procedures and tests of all medical imaging devices.
17. Organise multidisciplinary courses for other HCP to increase MP awareness and knowledge.
18. Set up a Failure Modes and Effects Analysis (FMEA) chart for every imaging modality to help prevent failures while targetting the causes and effects of the failures. This forecasting technique is a safety assessment that ensures past errors or mistakes are not repeated.
19. Set up a Threat Analysis chart of all the threats that the team faces with regards to the constancy testing programme.

IV. Conclusions

The MP team is blessed with inherent strengths based on the MPPs' physics/engineering background but the programme is not sufficiently developed because of issues arising mainly from MP being a fairly young and somewhat inexperienced profession in Malta. Thankfully, the current programme has higher ratings for S/O themes than W/T themes. Whilst relatively positive results have emerged, the programme may be further improved by developing a comprehensive strategic plan including a study of key success factors, analysis of competition from other professions, translating strategies into action plans and establishing accurate controls. This ensures that the conditions for greater success of the programme are delivered. Whilst SWOT analysis has been used extensively as a tool for improving strategies within organisations, nothing has been found in the literature regarding the use of SWOT for improving constancy testing programmes of medical imaging devices. This makes this study a first of its kind.

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A NOVEL TEST OBJECT AND METHOD FOR PRECISE ASSESSMENT OF GAMMA CAMERA INTRINSIC SPATIAL RESOLUTION DURING ACCEPTANCE TESTING

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Abstract. **Acceptance testing (AT) is considered a mandatory and particularly important procedure for any newly installed gamma camera because the parameters declared by the manufacturer are verified. Therefore, the AT measurement methods must produce results that are accurate and reproducible.**

Intrinsic spatial resolution (ISR) is one of the most important parameters of the gamma camera detector. Objective methods for the assessment of ISR are preferred because they provide quantitative assessment necessary for tracking the performance of a camera over time. Many authors emphasize the variability of ISR across the detector field. A commonly accepted method of quantification of ISR uses a parallel line equally spacing (PLES) phantom. More than 200 measurements are made on the image of the PLES phantom to compensate for the variability of ISR – approximately 100 in the central field of view (CFOV) and approximately 100 in the useful field of view (UFOV). The results for ISR in CFOV and UFOV are an average of the respective 100 measurements. The problem is that these results differ from the specification.

A new slit phantom - TP-phantom has been developed, that allows for an assessment of ISR at any point and in any direction of the detector field. Based on this TP-phantom's ability, a new method for objective, fast and accurate assessment of ISR is proposed. This method yields results that are reproducible and close to the specification of the gamma camera, which makes it particularly appropriate for acceptance testing. The need for an appropriate selection of reference points for ISR assessment is discussed.

An important advantage of TP-phantom is that it ensures zero radiation hazard for staff. A combination of 2 TP-phantoms can simulate a point source of very small size at a sufficiently high intensity.

Keywords - **phantom, acceptance testing, intrinsic resolution, gamma camera,**

I. INTRODUCTION

Acceptance testing (AT) is considered mandatory and particularly important for any newly installed gamma camera. On the one hand, the parameters declared by the manufacturer are verified until, the true, actual parameters of the apparatus are established, which will be the basis for monitoring the condition of the apparatus during its lifetime of clinical use. Therefore, the AT measurement methods must produce accurate and reproducible results.

The intrinsic spatial resolution (ISR) of a gamma camera is considered one of the most significant parameters of a gamma camera. The ISR may be measured either subjectively or objectively. Subjective methods rely on visual evaluation of transmission images of phantoms. The accuracy of subjective methods depends on both phantom design and the user's perceptivity and experience. Our preliminary study has shown why subjective methods are not suitable for AT of ISR [1].

Objective methods are preferred because they provide quantitative assessment necessary for tracking the performance of a camera over time. They (Objective measures) are based on the point or line spread function (LSF), ISR often being quoted as the full width at half maximum (FWHM). The generally accepted objective method of assessment of ISR uses parallel line equally spacing (PLES) phantom. The procedure with PLES phantom involves more than 200 measurements on the image of the PLES phantom – approximately 100 in CFOV and approximately 100 in UFOV. The result for ISR is quoted as an average of these100 measurements in CFOV and UFOV respectively. The reason to adopt such a complicated and time-consuming approach instead a single measurement is the variability of ISR across the detector field. This variability prevents an accurate assessment of the ISR.

The purpose of the current study was to develop a phantom and method for fast and accurate assessment of ISR during acceptance testing.

II. MATERIALS AND METHODS

At the core of the proposed practical solution to the problem with ISR assessment is the development of a test object (phantom) with flexible slit width. An additional purpose of the current study was to investigate the influence of the slit width on the accuracy of the assessment of the ISR. In conventional PLES phantom, the choice of a 1 mm slit width is a deliberate tradeoff between sensitivity (counting time) and accuracy of the FWHM measurement [2].

The developed slit phantom consisted of 2 lead plates 33x110x5 mm each, which were fixed stationary with 2 additional aluminum tiles (Fig. 1). The slit width between the two lead plates could be adjusted down to a size much smaller than 1 mm. The desired slit width was established

Fig. 1 Slit phantom with encapsulated source.

with an insert, which was gamma radiation transparent - for example, cardboard with corresponding thickness. The next step was to provide appropriate irradiation of the slit.

It is apparent that in the present case the traditional way of irradiation with a point source at a distance of 4-5 times the diameter of the field and closing the rest of the field with lead leaves was too inconvenient and not safe for scintillation crystal. That is why an encapsulated source – vial with 99mTc solution in a container was implemented (Fig. 1). At the bottom of the container, a hole with a diameter of 25 mm was cut. The slit of the phantom was irradiated through this hole. The container fully shields the vial from the environment and provides zero radiation hazard for staff. For convenience, the combination of slit phantom and encapsulated source in a container with an opening at the bottom would be referred to as TP-phantom. While working with the TP-phantom count rate was adjusted by changing the activity in the vial or by placing copper plates under the vial.

The gamma cameras taking part in this study were Philips Meridian, Siemens Intevo, Siemens E.CAM, and GE Millennium. All of them declared intrinsic resolution of 3.8 mm in CFOV and 3.9 mm in UFOV.

All measurements were made under fully calibrated detector and measurement conditions according to the National Electrical Manufacturers Association (NEMA) [2] and American Association of Physics in Medicine (AAMP) [3]: count rate between 10 and 14 kcps, peak of the LSF profile - above 2 000 counts, acquisition matrix 1024x1024. The energy window for 99mTc was the one recommended for clinical use.

The TP-phantom provided an excellent opportunity to easily determine the correlation between slit width and ISR measurement results. Measurements were made in the fixed place on the crystal as slit width was changed from 0.2 mm to 1.2 mm with a step of 0.2 mm. Fig. 2 shows a progressive improvement in ISR result with a decrease in slit width. For

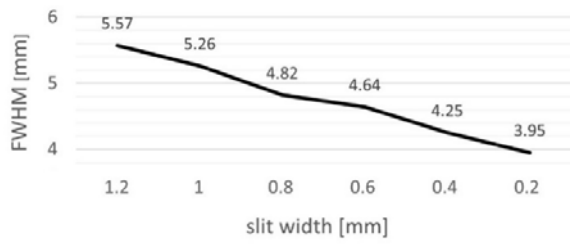

Fig. 2 Effect of slit width on spatial resolution result.

subsequent measurements, a 0.2 mm slot width was established which in camera resolution 3,8 mm meets the requirement [2]:

Camera resolution / slit width > 10

Which, with an expected camera resolution of 3.8 mm yields

3,8 / 0,2 = 19.

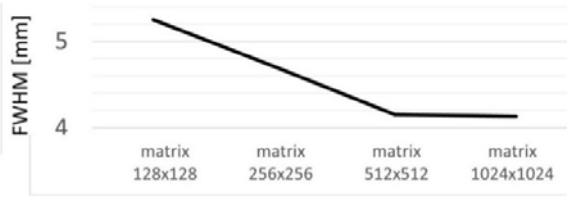

Fig. 3 Effect of pixel size on spatial resolution result

The next step was to determine the impact of the pixel size of the acquisition matrix on the accuracy of the measured spatial resolution. The TP-phantom with 0.2 mm slit width was placed stationary on the crystal and four acquisitions were made at the different sizes of the recording matrix. The result shown in Fig 3 shows that it is advisable to work with matrix 1024x1024. Fortunately, matrix 1024x1024 has been available in all gamma cameras for the last 10 or more years. The pixel size of this matrix is close to 0.52 mm at LFOV cameras and meets the second condition requested [2,3]:

Pixel size / FWHM < 0.2

Which, with an expected camera resolution of 3.8 mm yields

0.52 / 3.8 = 0.14

What remained to be studied was the third condition on which the accuracy of camera ISR measurement depends.

In routine quality control of working with the TP-phantom in the past years, it was very difficult to interpret the paradoxical fact that sometimes spatial resolution in UFOV is better than spatial resolution in CFOV. To find the possible cause of this effect, spatial resolution was measured with the TP-phantom along the small axis of the detector with step 1 cm. The line of measurement included both parts of the field – CFOV and UFOV (Fig. 4). Measurements were only in one direction Y along the dashed line through the

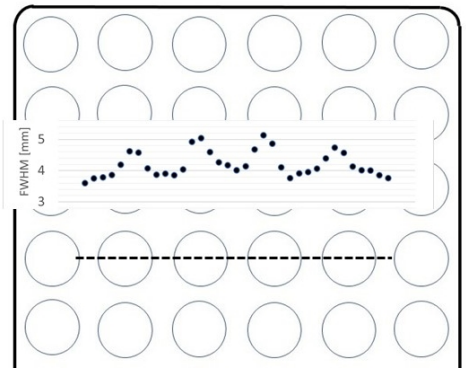

Fig. 4 Stepwise variation of spatial resolution along the line passing through the centers of the PMT

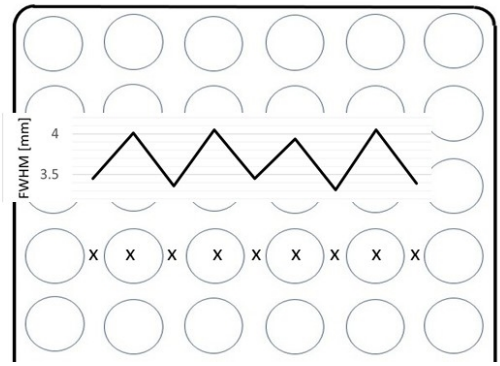

Fig. 5 Graph of spatial resolution at 9 reference points marked with "x" of PMT "matrix-stacking" detector.

PMT centers. There was no reason to expect that the graph of measurements at the same points on (in) the X direction will differ.

The results of the measurements in graphic form showed a change in a form similar to the sinusoid. After thoroughly reviewing relevant studies, we found that this phenomenon was published as early as 2006 [5] and later in 2009 [6], but it was not reflected in the ISR measurement procedure in official documents like NEMA [2] and AAPM [3]. Fig. 4's graph shows that there are specific places of minimum and maximum of ISR - the minimums are invariably at the limit between two PMTs - and the maximums coincide with PMT centers. The average of the minimums in Fig. 4 was 3.74 mm, while the average of the maximums - 4.73 mm was 26.5% higher. The declared ISR for this camera was 3.8 mm in CFOV, which practically corresponded to the average of the minimums.

The main conclusion of this example was that in the AT procedure of ISR, in addition to the requirements set out above for slit width of the line source and pixel size of the acquisition matrix, a third condition - appropriate permanent locations for ISR measurement - which we hereinafter refer to as reference points - must also be included.

Fig. 4's graph unequivocally shows that ISR has clearly localized maximums on the PMT center and minimums halfway between two PMT centers. This is a reason to use these points with a clear fixed location as reference points during acceptance testing and for the annual survey as well to track the state of ISR over time.

The following paragraphs propose sample options for selecting reference points for the objective assessment of ISR as an indicator of the detector's performance.

By arranging PMT in a rectangular detector field, gamma cameras can be conditionally divided into 2 types – those where PMT are arranged in the type "matrix" and those in which PMT are arranged in "chess-board" order.

On a "matrix" type PMT detector measurements of ISR were made (in Y direction only) in a series of 9 points marked with an "x". The results in graphic form are presented in Fig. 5. The average of the minimums was 3.9 mm while the average of the maximums was 4.74 mm. The difference between the max average and the min average was 21%. Apparently, the ISR value in the specification matched the average value of the minimums.

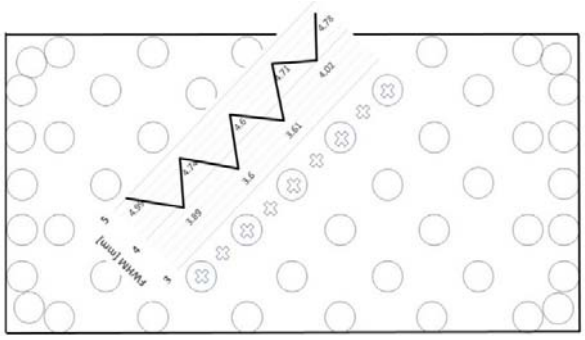

Fig. 6 Graphs of spatial resolution at 9 reference points marked with "x" in chess-board" arrangement of PMT.

In the case of the "chess-board" order of PMT measurements were made on a series of 9 points marked with an "x" (Fig. 6) – 5 measurements on the PMT and 4 measurements half the distance between two adjacent PMT. The selected PMT line passed through the center of the detector field. The average of the minimums was 3.9 mm while the average of the maximums was 4.74 mm. The difference between the max average and the min average was 20.5%. Apparently, the average value of the minimums matched the ISR value in the specification.

A possible extension of the functionality of the TP-phantom is adding a second slit phantom rotated at 90°. This configuration simulates a well-collimated point source with

a very small diameter (0.2 mm in this case). This way a point source of very small size can be realized which is difficult or almost impossible to achieve by drilling a physical hole into a lead plate.

This configuration can be used in a small-field-of-view gamma camera by reducing the size of the TP-phantom - a smaller plastic vessel for 99mTc solution with appropriate shielding and corresponding downsized slit phantom.

WARNING. Intrinsic measurement involves the removal of the collimator with the attendant risk to the crystal. Care must be taken so as to not damage the crystal. The phantom's small dimensions allow it to position itself at any point of UFOV, which is usually done by sliding. Sliding the phantom onto the easily damaged surface of the aluminum foil which covers the scintillation crystal poses a high risk. Therefore, it is strongly recommended to cover the scintillation crystal with transparent foil (1-2 mm) immediately after removing the collimator and then place the TP-phantom on the foil. Additionally, it is recommended to place a pad of soft fabric under the TP-phantom.

III. Discussion

An example of the inhomogeneous distribution of ISR in the detector field, which is always in front of our eyes, is the image of a 4-quadrant phantom with a high resolution starting at 2 mm bar width (Fig. 7). At an appropriate setting of the brightness window blurring of the bars above PMT becomes visible. Apparently, the lower resolution above PMT causes the lines in the quadrant with the narrowest bars to be displayed blurry. Note that blur is only in the areas around the PMT center, while in areas between PMT ISR is partially preserved, which confirms the results of this study.

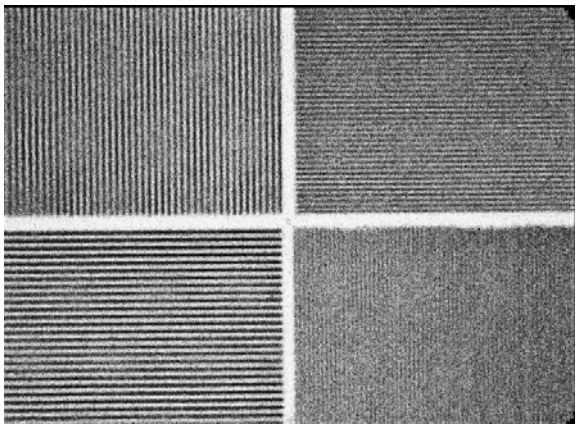

Fig. 7 Transmission image of four quadrant bar phantom.

The variation of the ISR across the detector field has been noticed on the first attempts at its assessment [4, 7]. The different resolution values along PLES phantom slits also confirm the variability of ISR on the detector field. It is therefore accepted to average the assessment of spatial resolution in an interval of 30 mm, which is further replicated in several (not specified exactly how many and where) locations [6]. The additional averaging of the result of X- and Y-direction measurements does not appear to give a consistent result. The reason is that it is made with two PLES phantoms, where the X and Y measurement points do not match.

Fig. 4‘s graph shows that in the interval between minimum and maximum, the intermediate resolution values can be found. In this sense, it can be expected that in a three-dimensional representation, the distribution of ISR across the detector plane will represent a geographical plain with "peaks" (PMTs) and "valleys" between them. It's easy to "get lost" in this huge set of different resolution values, which justifies the need for fixed permanent points of ISR measurement. Moreover - It becomes clear that the division of the detector field of CFOV and UFOV is conditional since "good" and "bad" ISR values can be found in both CFOV and UFOV. Most likely the ISR parameter in gamma camera specification is an average of the lowest points of "valleys"(the minimums).

We can take advantage of the opportunities provided by the TP-phantom to determine objectively and relatively quickly ISR. For PMT “matrix-stacking” detectors (Fig. 5), the average of the intermediate 3 results is an assessment of ISR in the CFOV, while the average of the endmost 2 results is an assessment of ISR in the UFOV.

The same model can be applied to “chess-board” order detectors (Fig. 6) - of the resulting 4 minimums (3.89 mm,3.6 mm, 3.61 mm, and 4.02 mm) the average of the intermediate 2 results is an assessment of ISR in the CFOV, while the average of the endmost 2 results is an assessment of ISR in the UFOV.

For a specific gamma camera, it becomes appropriate to determine a permanent row/s of PMTs to track the state of ISR over time.

TP-phantom allows measuring ISR at the same point in different directions with a simple rotation of the slit in the desired direction. This ability opens up an excellent opportunity to explore whether it is really necessary to measure ISR in both directions X and Y (as is the current rule) or it can be replaced, for example, by a single measurement with a slit of the TP- phantom rotated on 45°.

The reference point selection models mentioned so far are just an example of obtaining consistent ISR assessments - but they by no means limit the creation of other models based on different criteria.

PLES phantom has some disadvantages (weak points) such as:

- The slit width of 1 mm that does not meet the requirement FWHM/slit width > 10
- The slits have a fixed location and restrict access to only certain places on the detector field

- It is necessary to average LSF over 30 mm along the slit in order to reduce the impact of ISR variability alongside the slit
- PLES phantom irradiation time is relatively long to get information with good statistics
- Two PLES phantoms are needed to measure spatial resolution in X and Y directions
- Inevitable staff radiation exposure during preparation and irradiation of the PLES phantom.
- The software of some of the modern gamma cameras does not allow to form ROI with dimensions 55 x 30 mm to perform measurements with the PLES phantom (Siemens, GE)

The proposed new TP-phantom provides some significant advantages over the conventional PLES phantom, the most important of which are:

- Adjustable width of the slit to a very small size
- Well-shielded source in a container, which assures zero radiation hazard for staff
- Ability to provide the required count rate (below 20 kcps) with a change of source activity which facilitates and speeds up the measurement process.
- Short time to collect reliable information (LSF peak > 2000 counts) in a single measurement
- Ability to test ISR at any point of UFOV and in any desired direction with a simple rotation of the TP-phantom
- The TP-phantom is several times lighter than the standard PLES phantom and therefore reduces the risk of crystal damage while complying with recommended safe operation measures.

IV. Conclusions

A new phantom has been developed for rapid and easy quantitative measurement of ISR at any point of the detector field of a gamma camera - the TP-phantom. This confirms the findings of other authors [5, 6] – the presence of pronounced maximums and minimums of ISR, the locations of which in the detector field are unambiguously determined.

Based on this information, it is proposed to introduce an additional requirement to the ones prescribed by NEMA [2] - a specific measurement location of ISR. A criterion for selecting reference points in the detector field is proposed, which meets the requirements of the QC for the unambiguousness of ISR measurement conditions and ensures the replicability of the assessment.

The accuracy of the results obtained with the new TR phantom is confirmed by the fact that they comply with the specification.

The proposed assessment method facilitates acceptance testing, allows to compare objectively gamma cameras in terms of ISR, and tracks the detector's performance over time.

An essential advantage of the TP-phantom is the ability to establish a very small slit width (e.g. 0.1 mm), making it suitable for measuring ISR on a small field of view gamma cameras. An interesting option for the practice is the realization of a point source of radiation of small diameter adding a second slit phantom rotated at 90^{o}.

A practical advantage of the TP-phantom is that reducing the slit width does not result in a loss of time because the count rate can be adjusted with a corresponding change of the activity of 99mTc in the vial while observing the requirement count rate < 20 kcps. This makes measurement with TP-phantom quick and easy.

Last, but not least, working with the TP-phantom ensures zero radiation hazard for staff.

It is envisioned that the TP-phantom will provide fruitful avenues for other applications in the practice of medical physics.

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SELECTING A COMPUTED TOMOGRAPHY SCANNER

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Abstract: **The paper briefly describes the features to consider while purchasing a CT scanner.**

Keywords – **Computed Tomography, Multidetector CT, Radiation Therapy, PET-CT, Radiation Dose**.

I. Introduction

Medical physicists are often asked which computed tomography scanner to select from the scanners offered by several CT manufacturers. The answer is not straightforward and simple since there are many factors to be considered in the selection process. Generally, selection process can be broadly classified into various phases, such as the need assessment and specification phase, data collection phase and reviewing of manufacturer's data prior to selecting a particular imaging system. Once the image system is chosen, then the process continues with working out the details regarding construction and installation phase, acceptance testing phase, clinical applications phase and the beginning of routine use and the establishment of service contracts. The selection process greatly benefits with discussions among the radiologists, technologists, medical physicists, and administrators (1). The team approach method is optimum in selecting/identifying the best imaging system to match the need. The purpose of this article is to focus on the selection of a new CT scanner and key features to consider during the selection process.

Clearly identifying the goals and the needs for a CT scanner is important in the selection process. For example, need to select a ***general-purpose CT scanner*** or a ***special-purpose CT scanner*** or ***a CT scanner mostly used for CT Simulation scans in Radiation Therapy*** or ***a CT scanner adjacent to an upcoming Proton Therapy Center***.

A ***general-purpose CT scanner*** is one that is used for routine imaging of all anatomical regions and accommodate all types of patients and act as a workhorse in the radiology or in the emergency department. Such CT scanners should require minimal down time and accommodate all types of patients and therefore such scanner does not necessarily have to be the latest and the greatest generation but rather a stable scanner that has already demonstrated consistency for routine clinical use.

On the other hand, if the CT scanner is required for ***specific applications*** such as ***pediatric, neuro-radiological applications (perfusion), cardiac imaging***, ***hybrid imaging such as PET-CT, SPECT-CT, etc.,*** then the selection of special-purpose scanner should focus in identifying not only the scanners but also how much resources manufacturer can provide for specific applications.

The need for a CT scanner in the Radiation Therapy departments typically span over whether the CT scanner will be used primarily for CT simulation or the scanner need to have advanced CT applications such as dual energy CT capabilities desired by the new Proton Therapy Center.

One common theme applicable to all types of use is the capability of CT scanner to deliver good image quality with least amount of radiation dose (2). Often neglected aspect in selecting new CT scanner is the availability of service. Generally, CT scanners are quite stable and often requires minimal care, however, if it requires services, then it is critical to assess during selection, the availability of service. This is especially critical in LMIC (Low to Middle Income Countries) since the CT scanner downtime can be lengthy and disruptive to the clinical service.

If a center is planning to install a new CT scanner, it is also critical to assess the siting of the scanner and evaluate the availability of desired electrical power strength in the building. In addition, it is also important to assess the radiation shielding required for the CT scanner room and it should be in compliance with regard to the existing regulations in the particular region or country.

A detailed article discussing the key features essential in the selection of the CT scanner, along with available resources will be published in the next issue of the IOMP Journal Medical Physics International (MPI, Dec 2022).

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TOWARDS POTENTIAL HARM ASSESSMENT FROM THE INDIVIDUAL PATIENT RADIATION DOSES IN IMAGING PROCEDURES: A PROPOSAL FOR A NEW QUANTITY

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Abstract — **Imaging procedures continue to advance rapidly and offer unprecedented benefits in health care. Even so, the potential harm from the associated radiation exposure has remained relevant and subject to strong public scrutiny. This necessitates a quantity to gauge this potential harm in such a way that it is reflective of the attributes of the patient, the imaging procedure, and the latest science on radiation effects. The current metrics fall short of such objectives, as they are either procedure-centric (not relatable across imaging modalities), or negligent of the patient attributes, such as size, sex and age that are known to strongly influence the potential harm. Without a relevant quantity, the (often minor) potential risk associated with imaging procedures cannot be reliably put into perspective with the (often significant) benefit from the procedures, nor can that potential be properly monitored, communicated, or researched.**

In this white paper, we propose a new quantity that alleviates some of the shortcomings of existing measures. The quantity, which may be termed potential radiation harm or detriment, builds upon the foundation of effective dose and its numerical quantification with additional inclusion of patient and exam attributes. The new quantity is devised to enhance the assessment, optimization, and communication related to medical imaging procedures, with potential for extension to other conditions or practices where individualizations of irradiation is needed.

I. Introduction

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II. Why should we quantify patient radiation dose in medical imaging?

There is a prevailing assumption in the scientific community, anchored to the de facto linear no threshold (LNT) model of stochastic radiation risk (NCRP 2018, ICRP 2021), that any radiation dose may involve a non-negligible likelihood of harm. This includes likelihood of harm to patients undergoing medical imaging. At low doses associated with the vast majority of imaging exams, this likelihood is small and stochastic. While the magnitude of this harm remains debatable, its likelihood cannot be dismissed. As patient safety is an integral mandate of healthcare – *First Do No Harm*, the very *likelihood* of harm necessitates a system by which it should be quantified, minimized, and put in perspective with the substantial benefit associated with medical imaging. Patients, families, and clinicians who care for them want to know – and do ask – radiological professionals for the magnitude of doses associated with their imaging exams and the associated risk. Stating there is no risk is not scientific and avoiding a proper quantification only leads to the presumption of higher risk than actual reality.

III. What have we used thus far to quantify patient dose in medical imaging?

Over the years, various quantities have been developed to gauge the magnitude of patient radiation dose in medical imaging. Common among them are those that reflect the standard radiation absorbed dose in a phantom associated with a particular imaging condition: e.g., CT Dose Index (CTDI) and Dose Length Product (DLP) for CT imaging (ICRP 2012). While practical, these metrics do not reflect the likelihood of harm to the patient and cannot be compared across modalities, clinical indications, or patients.

Alternatively, Effective Dose, typically expressed in millisievert, has been used as a way to evaluate individual patient dose in a way that is independent of the modality for specific imaging exams. Effective dose has been developed by the International Commission of Radiological Protection (ICRP) as a dose quantity with a link to risks of health detriment from stochastic effects, for quantifying occupational and public doses, with the main objective of exposure limitation and risk management (ICRP, 2007). However, it has become a common metric for quantifying patient radiation doses across populations and medical imaging modalities in practice and publications with over 20,000 publications in the last 10 years alone (Zhang 2012, Brindhaban 2020, Casiraghi 2021, Fu 2021a).

This use has in fact significantly enhanced the reach and prevalence of Effective Dose, well beyond its originally designed purpose. However, this use of Effective Dose is 'off-label'. Effective Dose was never intended to capture potential harm to individual patients as it intentionally averages the effect of age, sex, size, and genetic radiosensitivity. Its original definition in fact emphasized that use of Effective Dose not as a substitute for specific risk analysis for individual cancer types using organ/tissue doses. Further, Effective Dose is calculated as a whole-body exposure estimate whereas patient imaging exposures are almost always only to a part of the body. Moreover, the non-commissioned and unguided use or Effective Dose for patient examinations has led to different calculations and implementations of Effective Dose across medicine, causing major confusion and inconsistences – no two millisieverts are created equal!

IV. WHY WE NEED TO DEFINE A NEW QUANTITY FOR AN IMAGING PATIENT'S RADIATION DOSE?

The non-orthodox, unrepresentative, and variable application of Effective Dose for assigning patient radiation doses is not mal-intentioned; it is rather a consequence of a lack of clear guidance for a better alternative. Medical exposures remain by far the leading source of artificial radiation exposure in the world (UNSCEAR 2022). As the community of radiation scientists, we have the opportunity and the responsibility to define a quantity that can better gauge the radiation dose associated with medical imaging.

V. WHAT SHOULD BE THE KEY INGREDIENTS OF THIS NEW QUANTITY?

The reason for assessing imaging radiation dose in the first place is its potential for harm to the patient. This is the only way that an imaging examination can be properly optimized, any shared decision-making to proceed with medical imaging can be communicated ethically, and dose benchmarking and management can have validity. As such, a proper quantity should be reflective of patient harm taking into consideration the unique attributes of the patient that contribute to this likelihood of harm. Any potential harm takes place within an organ or tissue. Therefore, the quantity should be informed by doses within and across organs, similar to the approach used for Effective Dose. These doses across organs should likewise be reflective of the attributes of the individual patient.

The quantification should further take into consideration other risk factors, such as age, sex, and patient body habitus, factors that are well recognized to influence dose and radiation risk. The quantity should also accommodate other factors once their influence has been well documented, such as the genetic disposition to radiation risk and the non-uniformity of the distribution of radiation dose across organs. There has been substantial formational work in patient- and exam-specific organ dose estimation that can be adapted in defining a standard methodology for patient-specific organ dosimetry (Li 2011, Choi 2020, Peng 2020, Samei 2020, Fu 2021b). Integrated with known age and sex weighted radiation risk factors, a quantity can likewise be defined to capture a supra-organ metric of radiation dose in imaging (Ria 2021).

VI. IS USING RISK AND AN ASSOCIATED UNIT A GOOD APPROACH TO QUANTIFY IMAGING RADIATION DOSE?

A quantity to reflect imaging radiation dose should provide an improved estimate of the potential harm from a patient's associated exposure to radiation (Ria et al., 2021). One thus may wish to capture that harm in terms of risk or a risk index (e.g., the likelihood of a cancer in 20 years). This approach, while used by many including principal authors of this article, is not ideal on five grounds:

1) A risk by definition assumes a likelihood of harm within a population identical to the patient. No two patients are created equal; therefore the quantity becomes hypothetical and not *patient*-specific as intended.
2) The method implies, by the virtue of ascribing a likelihood of harm to the patient, too much certainty on the science of radiation biology – there are still many unknowns and we should be careful not to project unwarranted certainty.

3) Any likelihood of harm depends, in a large part, on many factors that the patient will experience *in the future*. Ascribing a futuristic likelihood of harm is speculative.

4) A likelihood of harm estimated for some decades later has little practical value when compared with often immediate likelihood of benefit that will come from the examination. The two likelihoods have dramatically *different perceived values due to the times scales* and cannot readily be compared or put in balance with one another (so-called discounting in economic theory).

5) A likelihood of harm in terms of quantitative assessments such as micromort qualifications (Howard 1980), are overly terrifying to many patients that may have difficulty understanding the stochastic nature of the harm, are already concerned with morbidity and mortality, and cannot readily differentiate between milli or micro qualifiers, e.g., 1000 of something is perceived as a big value regardless of the units.

VII. WHAT QUANTITY AND AN ASSOCIATED UNIT SHOULD BE THE GAUGE OF IMAGING PATIENT RADIATION DOSE?

It seems prudent that an ideal quantity should take advantage of the prevalence, familiarity, and quantitative values of similar magnitude to those of Effective Dose to facilitate its adoption. Such a quantity may also be relatable to potential radiation risk, if so desired, but not be a direct reflection of risk – per points above – echoing the philosophy that led to the definition of Effective Dose in the first place. To avoid confusion, we do not recommend the use of the term "effective dose" in the nomenclature for the new quantity.

Informed by the rationales detailed above, we suggest this quantity should be described in a more generic manner relating to potential harm or detriment from radiation exposure. We have in fact considered the term *potential radiation harm* as a possible candidate for such a quantity. Such a term, in addition to alleviating the limitations of existing alternative quantities, offers unique advantages:

1) A quantity characterized as a *harm* or detriment can encapsulate (if needed) other effects of radiation exposure beyond stochastic risk and cancer induction.

2) A quantity characterized as a *radiation* harm can reflect directly what the patient would understand, the burden of radiation, beyond technical terms such as risk or dose.

3) A quantity characterized as a *potential* radiation harm can more authentically reflect the state of the underlying science of harm from radiation exposure, not all of which is fully known.

4) A quantity based on the best available data on radiation risk, will provide a more science-based quantity than effective dose, which used approximate weighting factors to facilitate simple calculations. This will allow more realistic and valid assessments of uncertainties in the characterization of radiation harm.

VIII. HOW CAN *POTENTIAL RADIATION HARM* BE DEFINED?

We propose the new quantity to follow the general framework of Effective Dose with the additional inclusion of patient and exam attributes. This approach echoes the formulation of ICRP Publication 147 (ICRP, 2021) to incorporate "approximate indicator of possible risk." The definition is based on estimation of organ doses and the exact irradiation condition of the patient. The approach follows these broad steps, exemplified for CT imaging but meant to extend to other modalities:

1) Modeling the patient geometry into a virtual form that captures the body habitus and organ locations of the patient. This will be done through matching patients to phantom libraries initially, but in time it should be possible to assess the components of the radiation dose to organs within the scan field directly (Choi et al., 2020, Fu et al., 2021).

2) Estimating dose to the individual organs of the patient via Monte Carlo-based methods or their derivatives taking into account all available information about the irradiation condition of the exam.

3) Estimating an overall level of radiation risk by summing individual organ doses multiplied by x_n-factored sensitivities, where x_n reflects numerous known radiation sensitivities (e.g., starting with age and sex, with additional personal factors such as smoking and patient size, progressively considered in the future extension of the methodology).

4) Scaling the estimated risk such that its numerical value matches to that of conventional Effective Dose (delivered to a 35 years-old adult) when the method is applied to the Effective Dose standard anatomical model and irradiation. In that way, the unit of the new quantity will not be sievert, but the quantity will have values in the same order of magnitude and scale as sievert.

IX. SHALL WE EXTEND THIS INDIVIDUAL QUANTITY TO WORKERS AND THE PUBLIC?

We propose this quantity to be applied initially to medical imaging, where a new quantity and guidelines are of current need (Ruehm 2022). However, the new quantity may as well be extended beyond patient imaging. Case in point, there are already individual dose limits in use, based on Effective Dose, set separately for female and for male astronauts (NAS 2021). Thus, there is a rationale to upgrade all such efforts to the new quantity.

X. WHAT ARE THE CRUCIAL REQUIREMENTS TO ENABLE THE CHARACTERIZATION OF *POTENTIAL RADIATION HARM*?

To enable the medical imaging community to compute and use a new measure of potential radiation harm proficiently and practically, we encourage a process to explicitly meet the following requirements:

1) Accuracy in modeling the patient
2) Accuracy in modeling the irradiation condition – first applied to CT imaging, and then to other modalities including nuclear medicine
3) Standardized description of the methodologies deployed
4) Benchmarking process
5) Incorporation of uncertainty in the quantity and its derivation (e.g., confidence interval)
6) Practical approximation strategies that accommodate resource-limited countries and settings

XI. CONCLUSIONS

The existing measures to gauge the potential radiation harm associated with medical imaging are inadequate to provide a quantitative account that is patient-relevant, technology-agnostic, and reflective of known factors of radiation risk. Currently the best candidate quantity for this purpose, Effective Dose, is a relatively poor discriminator, despite the considerable efforts to convey the link to stochastic effects. Consequently, efforts to justify and optimize medical imaging procedures and to communicate regarding benefit/risk are negatively influenced. In this article, we offer a proposal for a new quantity and metrology with the hope to enhance the assessment, optimization, and communication about medical imaging exposures to the benefit of all patients and the practices.

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PROFESSOR EHSAN SAMEI AWARDED THE MARIE SKLODOWSKA-CURIE AWARD OF IOMP, 2022

The Marie Sklodowska-Curie Award is one of the highest awards given by the International Organization for Medical Physics. It was established to honour scientists who have distinguished themselves by their contributions to education and training, advancement of medical physics knowledge based upon independent original research and/or advancement of the medical physics profession.

Ehsan Samei is the Reed and Martha Rice Distinguished Professor of Radiology, and Professor of Medical Physics, Biomedical Engineering, Physics, and Electrical and Computer Engineering at Duke University, where he also serves as Chief Imaging Physicist, Director of the Ravin Advanced Imaging Laboratories, and Director of the Center for Virtual Imaging Trials. He completed his graduate studies at Georgia Tech and the University of Michigan, and post-graduate training at Henry Ford Hospital. He has been a visionary leader and founding force behind major scientific, educational, and professional initiatives in medical physics, including the Society of Directors of Academic Medical Physics Programs, the Center for Virtual Imaging Trials, the Medical Physics 3.0 initiative, the Duke Medical Physics Graduate, Residency, and the Clinical Imaging Physics units.

Professor Samei's research ranges from his pioneering work on medical displays, now the basis of international standards in that area, to formative work on in-phantom and in vivo image quality assessment, dosimetry and dose-quality monitoring and optimization, and virtual clinical trials. Most notably, he has been a visionary role model for translational medical physics, where the clinical utility of innovations is measured not in terms of publications alone but in terms of patient-informed practice of physics in clinical medicine. This has been the primary motivation behind the Medical Physics 3.0 movement, transforming the field towards broad(er) application of physics in medicine and more intentional commitment to integrating science and relevance in medical physics scholarship. This is echoed through his over 330 refereed papers, 4 books, and hundreds of invited lectures and conference papers. He has also been a committed educator, mentoring over 135 graduate students and junior scientists.

Professor Samei has received numerous extramural grants, been recognized as a Distinguished Investigator by the Academy of Radiology Research, and awarded the Farrington Daniels Award by the AAPM, Jimmy O. Fenn Lifetime Achievement Award by the SEAAPM, and fellowships of the AAPM, SPIE, AIMBE, ACR, and IOMP. He is a diplomate of the American Board of Radiology. Throughout his career, he has faithfully served major professional organizations, journals, and grant agencies, and has held elected leadership positions including the presidency of the SDAMPP, the chairmanship of the SPIE International Medical Imaging Symposium, and the upcoming presidency of the AAPM.

We in IOMP are delighted and honored to have Ehsan as the 2022 Marie Sklodowska-Curie Awardee.

Heartiest Congratulations on behalf of IOMP Executive Committee.

Madan M. Rehani, PhD
President, IOMP

PROFESSOR GEORGE STARKSCHALL AWARDED THE HAROLD JOHNS MEDAL, 2022

George Starkschall is Professor Emeritus in the Department of Radiation Physics at the University of Texas MD Anderson Cancer Center, at which he first arrived in 1985. However, since 2010, he has served as the Executive Secretary of the Commission on Accreditation of Medical Physics Educational Programs, known as CAMPEP. This is a fitting position for George, as he has held a long-standing interest in education, particularly that of medical physicists.

Immediately after arriving at MD Anderson, George was appointed to the Radiation Oncology Residency Training Committee. He was subsequently appointed to the Curriculum Committee of the Graduate School of the Biomedical Sciences, and to the Medical Physics Graduate Program Steering Committee. He went on to Chair the Steering Committee, and served on, or chaired numerous other committees related to education at MD Anderson. A key position was as Director of Education in the Department of Radiation Physics, during which he was also Director of the Medical Physics Graduate Program.

As Director of Education, George introduced several teaching innovations into the medical physics graduate program. Chief among these were Peer Instruction, Flipped Learning, and Problem-Based Instruction. These techniques were substantial improvements on the conventional lectures offered by many educators, and George's success prompted several of his colleagues to emulate his style.

In addition to his passion for education, George put his skills and interests in programming to good use. He contributed to the development of one of the first clinical three-dimensional treatment planning systems, a radiotherapy picture archival and communication system, and an early method for obtaining time-dependent computed tomography image data sets.

During this time George was promoted to full Professor in the Department of Radiation Physics. He served in that position until his formal retirement in 2010, after which he returned in a part time capacity as a Research Professor and Distinguished Senior Lecturer. This enabled him to continue to contribute to the Department's education programs and allowed him focus on the quality and standardization of education.

It was at this time that George began his work with CAMPEP, which has set standards for quality for graduate and residency training programs. Accreditation by CAMPEP is now effectively a requirement in the U.S. and assures that graduates meet the requirements for medical physicist certification by the American Board of Radiology.

Between them, George and his wife Frada Boxer have four children and five grandchildren. They divide their time between Houston and the mountains of southwestern Colorado.

The IOMP is delighted and honored and to name George as the recipient of the 2022 Harold Johns Medal.

Congratulations on behalf of the IOMP Executive Committee.

Prof. Geoff Ibbott
Chair IOMP Scientific Committee

PROFESSOR XIE GEORGE XU AWARDED THE IUPESM AWARD OF MERIT, 2022

Prof. Xie George Xu is currently professor of Nuclear Science and Radiation Oncology, and director of Institute of Nuclear Medical Physics, the University of Science and Technology of China (Hefei, China). Before relocating to China recently, he spent 25 years at Rensselaer Polytechnic Institute (Troy, New York, USA) where his academic ranks included the Edward E. Hood Endowed Chair Professor of Engineering. He received a Ph. D. in Nuclear Engineering (health/medical physics focus) from Texas A&M University (College Station, Texas, USA) in 1994. Since 1995, Prof. Xu has mentored nearly 100 Ph.D and M.S. students in U.S. and China.

His research has dealt with "radiation dosimetry" for a wide range of challenges in radiation protection, medical imaging, and radiotherapy applications. His publication list includes 2 books, 220 peer-reviewed papers/chapters, 430 abstracts, and 150 invited talks. Widely known for his work on "computational phantoms" and "advanced Monte Carlo simulations", Prof. Xu is a fellow of American Nuclear Society (ANS), Health Physics Society (HPS), American Association of Physicists in Medicine (AAPM) and American Institute for Medical and Biological Engineering (AIMBE), as well as a council member of the National Council on Radiation Protection and Measurement (NCRP) and a past president of the Council on Ionizing Radiation Measurements and Standards (CIRMS). For two decades, he served on the editorial board of Medical Physics and Physics in Medicine & Biology. Prof. Xu has received numerous awards including most notably NSF's CAREER Award, CIRMS Randal S. Caswell Award for Distinguished Achievements (2015), HPS Distinguished Scientific Achievement Award (2018), ANS Arthur Holly Compton Award in Education (2020), ANS Rockwell Lifetime Achievement Award in Radiation Protection and Shielding (2020), AAPM Edith H. Quimby Award for Lifetime Achievement in Medical Physics (2020), and the IUPESM Award of Merit in Medical Physics (2022). Prof. Xu is the co-founder and president of Virtual Phantoms Inc. (The VirtualDose software - a CT and IR patient dose reporting software) and Wisdom Tech, Inc. (The ARCHER software - a GPU-based Monte Carlo dose computing software for treatment planning and dose QA verification).

We in IOMP are delighted and honored to have George as the awardee of the 2022 IUPESM Award of Merit In Medical Physics.

Heartiest Congratulations on behalf of IOMP Executive Committee.

Madan M. Rehani, PhD
President, IOMP

BOOK REVIEW

"INTRODUCTION TO MEDICAL PHYSICS", EDITED BY STEPHEN KEEVIL, RENATO PADOVANI, SLAVIK TABAKOV, TONY GREENER AND CORNELIUS LEWIS

De Vita, Enrico[1,2]

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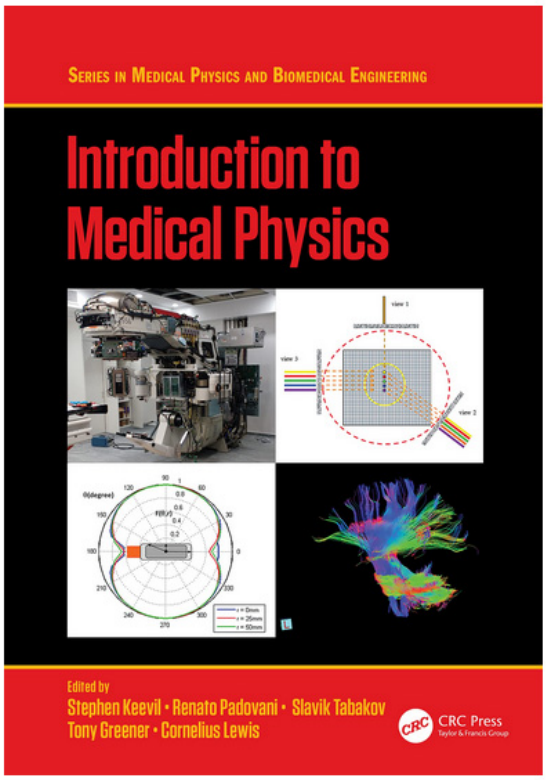

Abstract— **This article is a brief review of the textbook "Introduction to Medical Physics", 1st edition, edited by Stephen Keevil, Renato Padovani, Slavik Tabakov, Tony Greener, Cornelius Lewis, 2022, CRC Press, USA, ISBN 9781498744799**

The book "Introduction to Medical Physics" – 1st Edition is a recently published textbook in the Series in Medical Physics and Biomedical Engineering, the official book series of the International Organization for Medical Physics (IOMP).

The book "Introduction to Medical Physics" is edited by Stephen Keevil, Renato Padovani, Slavik Tabakov, Tony Greener, Cornelius Lewis. Many of the editors have experience in leading important Medical Physics or Medical Engineering and Physics Departments in the UK and Italy. Most importantly they have all been heavily involved in the delivery and continuous improvement of academic programmes in medical physics, in Europe as well as in low and middle- income countries, over the past few decades.

Whilst the global impact of medical physics and medical physicists in medical imaging and radiotherapy keeps increasing, and there is an increased need for trained physicist, few textbooks are available that comprehensively cover all of the field of medical physics. This book aims to cover this gap and provide a resource for undergraduate and graduate students aspiring to become medical physicists by presenting in a well-structured format the essential knowledge-base required for in-hospital training and subsequent practice.

The editors put together a team of practicing experts, comprising both clinical scientists and academics mostly from the UK and Italy, to cover each specialism in depth, with clear descriptions, explanations, supporting diagrams and figures.

The layout is in a single column. The total volume of the book is 500 pages, and there are 40 colour and 303 B/W illustrations. The book is comprised of 15 Chapters, each with its own Bibliography.

The book assumes some familiarity with physics and maths concepts that is usually acquired in the foundation year of physics, biomedical engineering or medical degrees.

Chapter one "Medical Physics, an Introduction" is authored by Perry Sprawls, medical physicist and Distinguished Emeritus Professor at Emory University has 11 main sections. It serves to introduce the role of Physics in Medicine, in diagnostic and therapeutic processes, and the role of medical physicists. It introduces the key imaging modalities and radiation therapy techniques, explains their individual differences and collective synergy. It is 13 pages long, supported by 13 figures.

Chapter two "Physics of Radiation Interaction and Dosimetry" is authored by editor Renato Padovani, Consultant at the International Centre for Theoretical Physics in Trieste and Charles Deehan, former Head of Radiotherapy at Leicester Royal Infirmary. It has 7 sections and 25 subsections over 33 pages, with 26 figures. Foundation chapter for radiotherapy and radiation protection, it describes the type of radiations used in

medicine, and introduces how they interact with tissue and the basics of dosimetry. Clear explanations are supported by clear diagrams and essential equations.

Chapter three "Ionising Radiation Detectors" is authored by Elizabeth Benson, now freelance consultant medical physicist in Radiation protection. It has 4 main sections: Modes of operation, Detector properties, Detector types, Detector applications. It provides a comprehensive description of detectors over 26 pages, with 32 subsections supported by 11 figures and 8 tables. The Detector Properties section is very usefully structured as questions-and-answers.

Chapter four "Biological Effects of Ionising Radiation" is authored by the editor Cornelius Lewis, previous Director of Medical Engineering and Physics at King's College Hospital, London, with Michele Avanzo, intraoperative radiation therapy physicist at Centro di Riferimento Oncologico in Aviano, Italy. It covers the topic in depth with 6 sections: Radiation Damage at a Cellular Level, Deterministic Effects, Stochastic Effects, Determining Stochastic risk, Risk Quantitation, The radiobiological basis of Radiotherapy. The chapter is 20 pages long with 28 sub-sections, supported with 10 figures, 3 tables and key equations.

Chapter five "Introduction to Diagnostic Radiology (X-Ray and Computed Tomography Imaging)" is edited by the editor Slavik Tabakov, Director of MSc in Medical Engineering and Physics at King's College London for over 18 years, with Paola Bregant, Medical Physicist at Azienda Ospedaliera Universitaria GI in Trieste. It has the following main sections: X-Ray Tube and Generator as a Source of Radiation; X-Ray Image Formation; X-Ray Imaging Methods and Their Application in Medicine; Image Quality in CR and FPD Systems; Computed Tomography (CT) Scanning. First of the key classic chapters, describing in detail X-ray production, X-ray imaging and CT, easy to read and understand with clear diagrams. The chapter is 48 pages long with 66 sub-sections, supported by 51 illustrations.

Chapter six "Nuclear Medicine Imaging" is authored by Elena De Ponti, and Luciano Bertocchi, Medical Physicists at Azienda Socio Sanitaria Territoriale, Monza, and Abdus Salam International Centre for Theoretical Physics, Trieste, respectively. It has 6 section parts: Nuclear Medicine Functional Imaging; Nuclear Decay Processes; Production of Unstable Isotopes; Radiopharmaceuticals; Gamma Camera Principles and Construction; Tomographic Acquisition. Another key chapter, also very well explained, provides the key concepts of nuclear medicine physics in a comprehensive yet concise style. The chapter is 29 pages long with 23 sub-sections, supported by 30 figures.

Chapter seven "Magnetic Resonance Imaging" is authored by the editor Stephen Keevil, Head of Medical Physics at Guy's and St Thomas Hospitals NHS Foundation Trust, London together with Renata Longo Professor of Medical Physics at University of Trieste. The 9 sections are: Historical overview, Nuclear Magnetic resonance, From Signals to Images: Encoding Spatial Information in NMR, k-space, Pulse Sequences and Contrast Manipulation, Artefacts: Problems, Solutions and New Ideas, Advanced Techniques, MRI Instrumentation, MRI Safety. A classic chapter, with a traditional structure, and clear understandable explanations. It provides a solid foundation for students and MRI users, enabling them to move on to tackle research MRI papers. The chapter is 62 pages long with 41 sub-sections, supported by 42 figures/diagrams.

Chapter eight "Ultrasound Imaging and Therapy" is authored by Raffaele Novario, Head of Medical Physics at Università degli Studi dell'Insubria, with Sabina Strocchi, Medical Physicist at ASST dei Sette Laghi, Varese. It has the following main sections: Interaction of Ultrasound and Tissue; Generation of Ultrasound; Diagnostic Modalities; Ultrasound Therapy. A concise and focused introduction to ultrasound in medicine. The chapter is 27 pages long with 20 sub-sections, supported by 10 figures and 27 equations.

Chapter nine "External Beam Radiotherapy" is authored by the editor Tony Greener, Head of Radiotherapy Physics, Guy's and St Thomas' NHS Foundation Trust, London, with Emma Jones and Christopher Thomas, Medical Physicists in the same Department. It has the following main parts: Beam Therapy Equioment, Clinical Dosimetry, Treatment Planning, Treatment Techniques, Imaging in External Beam Radiotherapy. First of 3 chapters on radiotherapy, covers all the physics of external beam radiotherapy with clarity through examples and many illustrative diagrams. The chapter is 60 pages long with 72 sub-sections, supported by 51 figures and 5 Tables.

Chapter ten "Brachytherapy" is authored by Mauro Carrara and Francesco Ziglio, Medical Physicists at Fondazione IRCCS Istituto Nazionale dei Tumori, and Santa Chiara Hospital, Trento, respectively. It has the following main sections: Delivery Systems and Applications; Radioactivity and Definitions; Radionuclides for HDR Brachytherapy; Gamma-Emitting Radionuclides for LDR Brachytherapy, Beta-Emitting Radionuclides for LDR Brachytherapy; Source Strength Measurement; Principles of Dose Distribution Calculation, Treatment Planning in Brachytherapy. The chapter is 33 pages long with 31 sub-sections, supported by 24 figures, 4 Tables, 16 equations and a comprehensive bibliography.

Chapter eleven "Molecular Radiotherapy" is authored by Lidia Strigari, Head of Medical Physics at Azienda Ospedaliero-Universitaria, of Bologna, Italy, and adjunct Professor. Similarly structured to the Brachytherapy chapter it has the following main parts: Delivery Strategy and Applications, Molecular Radiotherapy Targeting,

Measurement of Radioactivity, Principles of Dose Calculation, Treatment Planning Systems, Radiopharmaceutical Targeting for MRT, Treatment Regime Optimisation, Dose Effect Relationship, Radioprotection. The chapter is 27 pages long, with 29 sub-sections, supported by 5 figures, 16 equations and 60 References.

Chapter twelve "Optical and Laser Techniques", is authored by Elizabeth Benson and Fiammetta Fedele, Head of Non-ionising Radiation at Guy's and St Thomas' NHS Foundation Trust, London. Has 4 main sections: Optical Radiation in Medicine, Lasers, Non-laser sources, Optical Radiation Protection. Another very well written chapter, where theoretical background is linked seamlessly with practical information from active practitioners.

The chapter is 37 pages long with 38 sub-sections, supported by 21 diagrams/figures and 11 Tables.

Chapter thirteen "Ionising Radiation Protection", is authored by the editor Cornelius Lewis with Jim Thurston Head of Medical Physics and Clinical Engineering at Dorset County Hospital NHS Foundation Trust. It has 4 main sections: Risks of Ionising Radiation, Principles of Radiation Protection – Justification, Optimisation and Limitation, A Framework for Ionising Radiation Protection, Radiation Protection in a Medical Context, Radiation Protection for Healthcare Workers, Radiation Protection for Patients, Exposures in Pregnancy. Building on chapter 4, it provides a well organized introduction to radiation protection principles, regulations and practice. The chapter is 23 pages long with 27 sub-sections, supported by 9 figures/diagrams.

Chapter fourteen "Image Processing" is authored by Dr Andrew King, Reader in Medical Image Analysis, in the School of Biomedical Engineering and Imaging Sciences at King's College London. A short chapter of 14 pages and 14 figures, it only covers Image Filtering and Image Segmentation; but does so in an effective way with clear examples and a case study, providing the readers with a good background for further exploration of this rapidly expanding field.

Chapter fifteen "Emerging Techniques" is authored by editors Michele Avanzo, Tony Greener and Slavik Tabakov, with Luigi Rigon from Trieste University, It has 3 main sections: Phase Contrast Imaging, Radiomics, Ultra-High Dose Rate Radiotherapy – FLASH-R. A very useful overview of these methods and their potential benefits for clinical applications. It has 16 pages, 28 sub-sections and 4 figures.

The extensive Index, over 12 pages makes it easy to find relevant material throughout the book.

"Introduction to Medical Physics", 1st edition, has been well structured by the editors with a coherent choice of topics. The language is academic yet focused, and each chapter provides a smooth learning curve from basic to more advanced.

As Director of the Medical Engineering and Physics MSc at King's College London for the past few years I have observed how learners' needs are changing. While students cherish the variety of opportunities that increasingly replace or complement traditional lectures/tutorials (workshops, practicals, visits, videos and online material) they also appear hungry for reliable introductory textbook material to consolidate their knowledge and review their learning.

I would like to congratulate editors and authors, as I believe they have clearly met their objectives; this textbook promises to be an excellent reference for Medical Physics trainees, MSc students in Medical Physics, and also highly valued by 3rd year BSc students. Medical specialists and other professionals with an interest in Medical Physics will also appreciate the comprehensive coverage and the clear and detailed explanations.

HISTORY

PAUL LANGEVIN (1872-1946): THE FATHER OF ULTRASONICS

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***Abstract*— The year 2022 marks the 150th anniversary of the birth of the French physicist Paul Langevin. In February 1917, Langevin invented the first piezoelectric ultrasound transducer, to be used as a means for detecting U-boats using ultrasonic echoes. This discovery opened the way for new scientific and practical investigations, eventually leading to the widespread use of ultrasound for medical imaging. Langevin is widely honoured in his native country, but there is a paucity of biographical material in English. In this article we present a translation of the obituary of Langevin by Frédéric Joliot-Curie, first published by the Royal Society of London in 1951. This translation allows a wider understanding of the life of this outstanding man of science and man of the people, whose fundamental contributions to ultrasonics have remained inadequately recognized.**

Keywords— **Ultrasound, Langevin, Joliot-Curie, Obituary, Medical.**

I. Introduction

The medical applications of ultrasound, diagnostic, therapeutic and surgical, are the most visible and tangible present-day evidence of the scientific work of the renowned French physicist, Paul Langevin. 2022 marks the 150th anniversary of his birth in Paris on 24 January 1872. In this article we present the first English translation of the obituary of Langevin by Frédéric Joliot-Curie, published in French by the Royal Society of London in 1951, five years after his death on 19 December 1946 [1]. Several biographies were published of Langevin in French during the years following his death [2,3,4]. His collected work has been republished [5,6]. There has even been a biography written in Russian [7]. But, apart from occasional more recent brief articles [8], there is a severe deficiency of publications in English about Langevin's extraordinary life and achievements. The following translation will begin to fill this gap.

The biography was signed 'F Joliot'. This French physicist is better known as Frédéric Joliot-Curie (1900-1958), who, with his wife, Marie Curie's daughter Irène, shared the 1935 Nobel Prize for chemistry for their discovery of artificial radioactivity. He had studied under Langevin at the *École supérieure de physique et chimie* in Paris before becoming Marie Curie's assistant. He led the celebration for Langevin's post-war return to Paris in 1945 [9].

Paul Langevin was a Foreign Member of the Royal Society. He received the Hughes Medal in 1915 for 'his important contributions to, and pre-eminent position in, electrical science' and the 1940 Copley Medal for 'his pioneer work on the electron theory of magnetism, his fundamental contributions to discharge of electricity in gases, and his important work in many branches of theoretical physics'. Neither citation mentions ultrasonics nor piezoelectricity. Recognition for his work in ultrasound did not develop until after his death, and remains muted even today.

Joliot's biography sets Langevin's work on ultrasound in the context of his life and other scientific work. In this context, the entry is brief: half a page, set between a summary of Langevin's contributions to the theory of relativity and an overview of his last work in which he developed a theory for the stopping power of fast neutrons, necessary for the design of nuclear reactors. Given the brevity of this entry, it is appropriate to add some more details of Langevin's work on ultrasonics and the piezo-electric properties of natural quartz crystals.

II. Langevin"s discovery

Langevin identified February 1917 as the date when he realised that the piezoelectric properties of quartz might be successfully exploited initially to receive and then to transmit ultrasound. He had already been working with the French Navy for two years to devise an ultrasonic system for the detection of enemy submarines. A Russian émigré, Constantin Chilowski, had pointed out that a practical directional beam of sound could be created if the frequency was high enough, a suggestion that had also been made by the British physicist Lewis Fry Richardson. By the end of 1916 the French team had designed a source of ultrasound at 100 kHz using a 'singing condenser', detecting the waves produced with a carbon granule microphone connected to a wireless receiver.

Langevin had been taught by Pierre Curie in his youth and, as he matured, he became part of an intimate quartet of Parisian physicists, with Marie and Pierre Curie and Jean Perrin. The Curie brothers, Jacques and Pierre, had demonstrated the piezoelectric properties of quartz in 1880 and had devised an instrument, the *quartz piézo-électrique*, with which Pierre and Marie had measured the radioactivity of radium. Langevin realised that this *quartz piézo-électrique* was cut along the wrong plane through the crystal to be efficient to detect ultrasound. A different slice orientation was required, so-called X-cut quartz, for which the electrical and strain axes were aligned. Still, it was not

until he tested a single crystal as a replacement for his carbon receiver that he knew that the piezoelectric properties, previously investigated only under static conditions, were retained at 100 kHz. Langevin delighted in this elegantly simple solution, an acoustic aerial that he described as 'a piece of stone, two plates of tinfoil' [10].

A further breakthrough was made later in 1917 by his discovery that the reciprocal nature of piezoelectricity could be exploited to generate ultrasound also. This opened the way to the design of the first pulse-echo transducer, which could be used both to emit a pulse of ultrasound, and to receive echoes. Langevin's discovery was disseminated as widely as possible, given the wartime conditions, to the Allied laboratories in Britain, USA and Italy. By the end of the war, successful ultrasonic pulse-echo systems were being tested by both French and British navies.

Joliot-Curie's obituary includes a pertinent comment about what happened next: "A new chapter in acoustics was opened and experiments .... were made possible at frequencies of several hundred millions per second. Many discoveries in physics and many applications in chemistry and biology have thus been made possible." Indeed, all post-war work on ultrasound derived directly or indirectly from Langevin's breakthrough. The Canadian Robert Boyle, who had led the British asdics team, and was now back in Alberta, investigated acoustic cavitation and ultrasonic metrology. The American Robert Wood had visited Langevin during the war and subsequently set up an ultrasound laboratory with Loomis at Tuxedo Park, New York demonstrating dramatic physical and biological effects. Similar work was carried out by Frank Lloyd Hopwood, physicist at St Bartholomew's Hospital Medical School, London, who had learned of Langevin's work through Boyle. Alexander Nicolson's development of the piezoelectric crystal Rochelle salt followed from the open communication with industrial engineers in the USA, as did Walter Cady's work on frequency control. Léon Brillouin predicted light scattering by ultrasonic waves. More details of these developments have been published in Medical Physics International [11] and elsewhere [12].

During the immediate post-war years, Langevin devoted much attention to the interpretation of Einstein's special theory of relativity [13], but maintained his consultancy with Toulon where he negotiated to establish a transducer laboratory [14]. 1923 marks the year when his wartime work on ultrasonics became public knowledge. In this year he presented an extensive course on ultrasonics at the *Collége de France*, the first of its kind anywhere in the world. He had a reputation as an outstanding teacher. His course included beam formation, acoustic shock formation, transducer design, transmission through layers and the physics of acoustic absorption. The contents were later written up and published by his student Pierre Biquard [15]. This led to further improvements in transducer design and in ultrasonic metrology [16] and supported the successful exploitation for civil echo-sounding and underwater navigation aids by Charles-Louis Florisson. Langevin's work with quartz transducers created the impetus for the development of other ultrasonic transducers for underwater applications using magnetostrictive devices and other piezoelectric materials. In spite of this, quartz remained the transducer material of choice in ultrasound laboratories for several decades.

III The link to medical ultrasound

There is an irony that ultrasound was being used for medical therapy in Germany by 1940, at the same time that Langevin himself was arrested by the Gestapo and subsequently held under house arrest in Troyes. Langevin did not live long enough to know about the first Congress on Ultrasound in Medicine, held in Erlangen in 1949, a congress almost entirely devoted to therapy [17]. His colleague Florisson, who attended the congress, reported that Langevin had predicted the potential therapeutic use of ultrasound. Indeed, there is evidence from a 1925 patent filed by Léon Brillouin that these friends had thought about the therapeutic rather than the destructive effects emphasised by others at this time [18]. However, it was left to the German physicist Reimar Pohlmann to establish its scientific rationale in 1939. By 1950 there were about a dozen manufacturers of therapeutic ultrasound equipment, and all but one of them were using quartz piezoelectric transducers, based on Langevin's discovery over thirty years earlier.

Joliot's biography was written as ultrasound was emerging as a therapeutic agent, but before pulse-echo detection had been investigated for medical diagnosis. By 1972, on the centenary of Langevin's birth, his student Pierre Biquard could finally add medical applications to 'the field of theoretical and experimental research and industrial applications' as outcomes directly attributable to Langevin's work on ultrasonics [19].

There are close similarities between X-rays and ultrasound in their historical passage from discovery to medical use. This is in spite of the obvious difference between the dramatic immediacy with which Roentgen's announcement was taken up by doctors and the long delay and slow subsequent development of ultrasound. Both men, Langevin and Roentgen, were eminent European physicists. Both immediately informed others of their discovery. In both cases, the breakthrough was based on earlier pure science: the investigation of the passage of an electric current through rarefied gasses resulted in the discovery of X-rays and the exploration of piezoelectricity led to practical ultrasound transducers. Both are examples of the enabling technology that led to subsequent developments. For both, the claim to originality was challenged, through Rutherford and Nicholson for piezoelectric transduction and by Lenard for x-rays. In both cases each technology went through an important change: from the gas tube to the Coolidge tube for x-rays: from quartz piezoelectric transducers to ferroelectric ceramic transducers for

ultrasound. Both new radiations stimulated fundamental new science, and were exploited in numerous scientific and industrial applications. In medicine, both radiations became used for both diagnosis and therapy. These discoveries form the rootstock of modern medical imaging. Paul Langevin should take his place alongside Wilhelm Roentgen as one of its founders.

ACKNOWLEDGMENT

We are indebted to the Royal Society for permission to publish the following translation of Frédéric Joliot-Curie's obituary of Paul Langevin, which first appeared in French in the Obituary Notices of Fellows of the Royal Society. We are also grateful for the assistance of Philippe Blondel in reviewing the accuracy of this translation.

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PAUL LANGEVIN 1872-1946

Obituary Notices of Fellows of the Royal Society, 1951;7:405–419

F Joliot

(Translated from the French)

Paul Langevin was born in Paris, in the Montmartre district, on January 23, 1872. His family was of very modest circumstances. His father, after having served as a Zouave in the Army[1], became a building surveyor. He family was from Falaise in Normandy. This small town in Calvados was also home to an Abbot Langevin.

Paul Langevin's mother, Marie-Adelaïde Pinel, was the grand-niece of a psychiatrist [2], very well known at the time and who had been a member of the Academy of Sciences in Paris.

He was attracted to scientific studies early in life and entered the Municipal School of Industrial Physics and Chemistry of the city of Paris where he became first of his class. This school had been very recently created - Paul Langevin was part of the seventh cohort - and it tried to inaugurate, in France, the teaching of physics and chemistry in which experimentation would hold a large part. In this School, which he was later to direct and where he made a profound impact, he was taught in particular by Pierre Curie. He was first in his class, and then moved to the *École normale supérieure,* which he attended between 1894 and 1897. And there, and also in the Physical Sciences *Agrégation* examination[3], he largely out-performed all his classmates, already affirming his exceptional qualities.

When we explore the unfolding of a life, however long and fruitful it may have been, we are almost always led to discover the occasion or the event that was decisive. For Paul Langevin, the granting of a scholarship abroad, in 1897, by the City of Paris, certainly had this character, for the young and already brilliant physicist was thus led to continue his studies at the Cavendish Laboratory in Cambridge. There he made the acquaintance of eminent physicists, who have since become famous, and he retained all his life a vivid memory from this period of his existence. His British colleagues had an impression that was summed up very well by Rutherford when he said to those of Langevin's pupils coming to Cavendish Laboratory – "Tell your Master that he will always be at home here".

1. [1] Possibly a conscript – Zoave were previously an élite infantry corps from N Africa
2. [2] Aliéniste
3. [3] The *Agrégation* is the highest level of French degree assessing suitability to teach a particular discipline at all levels, including higher education.

Back in France in 1900, Paul Langevin continued his research at the Sorbonne, where he was laboratory assistant and then *Chef de Travaux* [4] for the Chair of Physics. This work, initiated at the *École normale supérieure* and continued at Cambridge, focused on the properties of X-rays, discovered shortly before by Roentgen, and on the ionization of gases. On April 28, 1900, his first publication appeared in the *Bulletin de la Société Française de Physique* entitled 'On the ionization of gases'. It is also on this subject that in 1902 he defended his doctoral thesis. So, under the title 'Researches on ionized gases' he published the first overall, experimental and theoretical account of his research on gaseous ions.

In 1905 he succeeded Pierre Curie as professor at the *École de physique et de chimie*. In 1909 he became Director of Studies of this School, where he was overall Director from 1925 until his death. This was an important part of the work to which he devoted all his life, only interrupted by the period during which France suffered from the joint domination of the Nazis and their French accomplices. The latter removed Langevin from his post, while their masters put him under arrest and then under house arrest in Troyes.

But it was not only at the School of Physics and Chemistry that he taught physics with extraordinary mastery. In 1909 the assembly of the *Collège de France*, anxious to maintain at a very high level the teaching and the scientific effort of the establishment, asked Paul Langevin to succeed Mascart. Paul Langevin had already assisted Mascart at the *Collège de France* for seven years. The exceptional value of his work and the fruitfulness of his teaching naturally led the assembly to choose the young and already famous physicist.

Paul Langevin had the pleasure of becoming the close colleague of Marcel Brillouin, of whom he had been a student at the *École normale* first, then at the *Collège de France*. He liked to talk about his former master and to express to him on many occasions his gratitude and affection.

Brillouin was always proud of his pupil and, in 1945, now retired to a small town in France, he wrote him a very moving letter for his 73rd birthday, some passages of which I reproduce below.

'I often think, in my almost rural loneliness, of the years when I had the pleasure of examining you, for entry to the *École normale*, without being able to find a limit to the scope, the precision, or the clarity of the knowledge acquired by you at the School of Physics and Chemistry which you now direct, and of the conversations with Pierre Curie. When you went to the *École normale*, your mind had already matured, and I do not think I taught you much.

'What added most to your knowledge is the visit that you made to the University of Cambridge shortly after leaving the *École normale*. There you found masters whose teaching was quite different from that given to us in France in primary or secondary education. A mature mind like yours was required to make fairly significant changes in order to make good use of meeting eminent scholars like J. J. Thomson and Lamor, and laboratory colleagues who would soon become famous, like Rutherford, Wilson and Townsend.

Returning from Cambridge you were, in my opinion, the typical professor destined for the *Collège de France* and I had no difficulty in sharing this opinion with my father-in-law Mascart. In this chair, you were able to give the full extent of your ability to assimilate, to complete and to present with astonishing clarity the newest and unforeseen events, caused by the unexpected results of Michelson, the importance of the change of the Lorentz factor and, in its simplest form at first and most complete a few years later, of Einstein's theory of relativity. Others may have written books more quickly on these fine theories; it is from you, at the *Collège de France*, that all those who understand them well, and with clarity, have learned them either from your lessons, or in the meetings with frequent discussions each Tuesday and Friday in the large and uncomfortable physics lecture theatre.

'What didn't you add ?! What have you not sown, then, almost without knowing it, with the remarks suggested to you by the objections and, not least, the comments of some of the assistants?'

Paul Langevin was a member of a large number of Academies and Scientific Societies in many countries. In particular, he had the very high honour in 1928 to succeed H. A. Lorentz, as chairman of the Scientific Committee of the Solvay International Physics Institute. In this capacity he assumed the presidency of the Solvay Physics Councils of 1930 and 1933. He was subsequently a member of the 'Royal Society' and, since June 25, 1934, of the Academy of Sciences of the *Institut de France*. By his scientific work and his teaching of extraordinary fertility, Paul Langevin has greatly contributed to placing physics in a dominant place in the sciences. In experimental work, mark of his thought can be found in the great movement of research and ideas which has taken place since the end of the last century and which has led us to give a more exact understanding of Nature and a better representation of the phenomena which occur there. Experimental discoveries forced him, along with the great physicists of his time, to criticize and profoundly modify these notions, concerning fundamental notions of time, space, mechanics and the structure of matter and radiation.

History demonstrates how turbulent this period was, both from the point of view of science and that of social life. In this turmoil, of which these two aspects are only apparently independent, there are men who have known how to dominate events and hold high the torch of truth. Paul Langevin was one of them.

When he tried to place his work as a physicist in the evolution of Science he liked to say that he had successively lived through the great revolutionary calls of relativity and quanta that physics has experienced during

4. [4] *préparateur* and *chef de travaux* were scientific staff posts

the previous fifty years. In this regard, he evoked the difficulties of adapting the mind to new ways of questioning nature. But he was able to overcome these difficulties. In the theory of ions, in the study of dia- and paramagnetism, in the theory of electric and magnetic birefringence, his work is absolutely fundamental. The great movements of ideas created by relativity, quanta and wave mechanics find in Langevin not only a follower, not only a prestigious educator but also a major participant, and he established the famous law of equivalence between matter and energy independently of Einstein.

Paul Langevin has always been convinced of the need for a close and continuous link between pure science and technique, between the scientist and the practitioner so that the latter is informed as widely as possible, through the general culture, through the results obtained by the scientist and where reciprocally the scientist can know the problems posed in practice, with the filtration and the generalization necessary to the various stages of scientific organization, and also to profit from the increasingly powerful material means available to it. He provided a significant illustration of this by resolving in 1915 the problem of the production and detection of ultrasonic waves, thus placing in the hands of the Allies a weapon that proved, during the two wars, so effective in the fight against German submarines.

An exceptionally talented teacher, he taught generations of scientists at the *Collège de France*, the *École de physique et chimie* and the *École normale supérieure de jeunes filles*, by always presenting to them the living aspect of science, of the science that is created. He did not think it necessary to limit his pro-religious activity to these tasks. His universal mind and his precision of judgment enabled him to analyze social problems in depth. Paul Langevin did not want to be part of an elite of scholars detached from real life. It was through action, as a militant in the larger community of workers, that he concerned himself with social problems.

To have fought for peace, for international solidarity, for social justice, against racist theories, all this pointed to Langevin as a target of choice for the Nazis. Despite the risk, Paul Langevin returned to Paris in 1940 and in October of that same year he was arrested, thrown in prison, and then put under house arrest in Troyes. His daughter Hélène was deported and his son-in-law, physicist Jacques Solomon, a fervent Communist, was shot. All these misfortunes deeply wounded Paul Langevin without shaking for a single minute either his courage or his certainty of the final triumph of justice over barbarism.

As Louis de Broglie so rightly said at the *Académie des sciences de l'institut de France*, on December 15, 1947:

'He brought to his opinions such sincerity, such conviction, such a passionate love for justice and suffering humanity that his attitude inspired respect, even in those who did not share his views. He knew how to rise above all petty considerations to that height of thought where all men of goodwill can agree. '

After his exile in Switzerland, Paul Langevin returned to Paris after its liberation and took over the management of the *École de physique et chimie* and his chair at the *Collège de France*. At the same time he was responsible for chairing the commission that was to finalize a profound reform of education in France. His health was greatly shaken by the terrible moral and physical injuries that were inflicted on him by the treasonable government of Vichy and the Nazis. After an existence entirely devoted to the two causes which he considered inseparable, Science and Justice, after having worked much, struggled much, suffered much, after having found and experienced great joys, Paul Langevin died on December 19, 1946. His last words were again to give to those around him confidence in science and hope for an approaching era of justice and kindness.

THE SCIENTIFIC WORK OF PAUL LANGEVIN

Gaseous ions

Langevin's first researches were devoted to the problem of the movement of gaseous ions affected by an electric field. From the experimental form of the saturation curve he deduced the value of the recombination coefficient from the mobilities. The study of the recombination coefficient led him to demonstrate that under the very low pressures prevailing in the upper atmosphere, there exists equilibrium a high concentration of ions in equilibrium. The Heaviside layer is thus explained as well as the persistence of conductivity during the night, when the ionizing action of solar radiation has disappeared. The same cycle of studies lead him to discover in the atmosphere the existence of large ions, whose mobility is several million times smaller than that of ordinary ions and which are made up of water droplets, one hundredth of a micron in diameter, having captured the charge of an ordinary ion. He was also able to give a theory for the formation of two types of clouds: stratus, cumulus or nimbus at an altitude of less than two thousand meters and higher clouds, the cirrus clouds, at an altitude of around ten thousand meters.

Paul Langevin then returned to the theory of free paths in order to generalize and extend it, taking into account the law of probability according to which the paths are distributed between two collisions. He thus dealt in a general way with the problem of mobility and diffusion and established that the ions in gases are constituted by a single layer of molecules maintained by electrostatic attraction around a charged centre. In a study published jointly with J. J. Rey, he demonstrated that the explanation of the conductivity of gases by collisions from thermal agitation was not in accordance with experiment. We now know that cosmic radiation is the origin of this conductibility.

Brownian motion

P. Langevin provided a new justification for Einstein's formula by breaking down the action of a molecular

collision on a particle into two terms: one of the terms concerns the normal action and corresponds to the viscosity, the other term is irregular and gives rise to Brownian motion. These kinds of question led him, in collaboration with Jean Perrin, to deepen the meaning of the second principle of thermodynamics so as to underline its statistical character, and to allow for the possibility of limitation (variability).

Electromagnetism

Besides studies on the mass of the electron and the variation of it with speed, Paul Langevin showed that the traditional theory of electromagnetic radiation completely interprets the phenomena of diffusion of light by through fluids, for example. A justification of the blue sky theory can be made in this way if a representation of molecules having an electric anisotropy is introduced. The various physicists who have studied these questions have made systematic use of this model, often referred to as the 'Langevin molecule'. Rayleigh's theory was thus integrated into electromagnetic theory.

Dia and paramagnetism

Paul Langevin made great progress in the theory of magnetism. His first publication on this subject dates from 1905. Paul Langevin took up Ampère's idea of the existence of molecular currents on a microscopic scale in connection with magnetic phenomena, but by introducing the recently-discovered electron. He thus succeeded in developing a theory of diamagnetism and paramagnetism.

Paul Langevin assumed that electrons follow closed orbits inside atoms. If e is the elementary charge, S the area of the swept orbit in time, T the magnetic moment, then M will be $M=eS/T$. The interpretation of diamagnetic phenomena results from assuming that, in such substances, the geometric sum of the magnetic moments is zero. If the whole is subjected to the action of an external magnetic field, the various electrons have their trajectories modified, the new movement being that which originally existed but *vis-à-vis* a system of axes revolving around the direction of magnetic field H, with the Larmor angular velocity $\omega = He/2m$. It follows that the atom takes an additional magnetic moment, directed in the opposite direction of the magnetic field and proportional to it.

As a first approximation, diamagnetism must represent atomic character and the constant associated with a molecule must be the sum of the atomic constants. This property is independent of temperature and the atomic constant and is, fixed by the number of electrons and the atomic dimensions, to an order of magnitude. Agreement with experiment is excellent and this interpretation of diamagnetism is universally accepted. Paul Langevin interpreted paramagnetism in terms of the orientation of atoms possessing a magnetic moment under the action of an external magnetic field.

The application of Boltzmann's law relating to the static distribution of such a set of atoms makes it possible to identify the following consequences:

(a) The magnetic moment and hence the paramagnetic susceptibility will depend on the absolute temperature and will vary inversely with it (a law discovered experimentally by P. Curie).

(b) Under the influence of a magnetic field of increasing magnitude a saturation of the developed magnetic moment is gradually established. The resulting relationship, known as Langevin's formula, has been found to be in very good agreement with experiment for various bodies, in particular gadolinium sulphate.

This theory made it possible to predict a remarkable phenomenon, which has been used for obtaining very low temperatures. The paramagnetic orientation must be accompanied by a rise in temperature and reciprocally an adiabatic demagnetization must result in a lowering of the temperature. By taking advantage of this latter property, de Haas was able to obtain the lowest temperatures that we know how to achieve.

The theory of the orientation of paramagnetic molecules under the influence of the magnetic field was to serve as a model, a short time later, for the dielectric theory developed by P. Debye.

Electric and magnetic birefringence

Paul Langevin applied the method of radiation that had been so successful for magnetism to give an explanation of the phenomena of electric and magnetic birefringence.

Cotton and Mouton attributed these effects to the orienting action of the fields on molecules exhibiting optical anisotropy as well as electrical or magnetic anisotropy. Using Boltzmann's law, Langevin was able to fully account for the variation of this effect with the strength of the field and with absolute temperature. This work on electric and magnetic birefringence has been the basis of much theoretical and experimental research activity on this subject.

Relativity

The negative results of the Michelson's experiments of optics and electromagnetism, undertaken to demonstrate the movement of the earth in relation to the ether or to the absolute space that it supports, could be interpreted by the contraction hypothesis of Fitzgerald and Lorentz in a satisfactory manner, without calling into question the notion of time. Paul Langevin showed that the same contraction hypothesis allowed a correct interpretation of the negative result of the Trouton and Noble experiment. Paul Langevin thus focused on the fundamental question of the relationship between mechanics and kinematics.

Mechanics, in the nineteenth century, seemed to have reached a perfect state of development, thanks to the successes achieved in astronomy. It seemed that the whole of the exact sciences were to be modeled on Newton's mechanics. These were supported by *a priori* ideas such as

the concept of mass and absolute time as well as a formal distinction between the concept of mass and that of energy.

It was to the great credit of Paul Langevin to have powerfully established the interdependence between concepts considered to be distinct *a priori* and of having shown that mechanics is only a branch of physics from which it should never have separated. If one associates the principle of conservation of energy with Lorentz-Einstein kinematics, required by the laws of electromagnetism, one can fully justify the dynamics of relativity. In this new dynamic, the separation established between the concepts of mass and energy is abolished. Paul Langevin's contribution to these developments was very important; he was able to justify, by different and more general reasoning, several of Einstein's results.

Paul Langevin treated various problems relating to relativistic mechanics and in particular the discussion of Sagnac's experiment, the interpretation of the deviations of atomic masses from multiples of the mass of the hydrogen atom, etc.

By various routes, all different from those followed by Einstein, Paul Langevin obtained the new dynamic that corresponds to a new space-time and gave conclusions regarding the inertia of energy and its consequences. Among these, there was one of extreme importance: that any change in the internal energy of a system results in a change in mass, obtained by dividing the change in energy by the square of the speed of the light.

From 1913, Paul Langevin gave remarkable confirmation of this relationship by interpreting, from the inertia of the energy, the differences between atomic masses and the integer multiples of the mass of hydrogen. From the discovery of nuclear reactions, Paul Langevin quantitatively related the energies released and the variations in mass of reacting nuclei.

Ultrasonic waves

During the 1914-1918 war, Paul Langevin was led to consider a problem that maritime disasters had already posed and for which the submarine warfare carried out by the Germans made the solution very urgent: the problem of detecting submarine obstacles. C. Chilowski had suggested that a directed sound beam could be used for this purpose, using very high frequency sounds (ultrasound) so that the dimensions of the emitting source need not be too large. To transform electromagnetic vibrations into acoustic vibrations, Paul Langevin had the idea of using the piezo-electric properties of quartz discovered by Pierre and Jacques Curie. The use of the elastic resonance of quartz, and the theory of the vibration of quartz-steel assemblies, allowed the practical design of ultrasonic sounders that played a great role in the two world wars.

At the same time, a new chapter in acoustics was opened and the experiments which had been only carried out on elastic vibrations of frequencies less than about twenty thousand were made possible at frequencies of several hundred millions per second. Many discoveries in physics and many applications in chemistry and biology have thus been made possible.

On collisions between neutrons and nuclei of any mass

Shortly before the military defeat of France in June 1940, Paul Langevin tackled a nuclear physics problem concerning collisions between fast neutrons and atomic nuclei of any mass. The solution to this difficult problem was a central part in the design of devices called uranium reactors in which the neutron moderator was composed of nuclei other than light hydrogen and absorbed very few thermal neutrons. The problems associated with these developments were being studied in France at that time. The calculation concerned the probability of a fast neutron passing through a retarding medium, slowing down by successive collisions with nuclei, to reach a kinetic energy between E and $E + dE$. This problem is quite simple to solve when the mass of (the atoms in) the retarding core is equal or substantially equal to that of the neutron, but it becomes very complicated when the masses are different. By a geometrically representative investigation of the required probabilities after one, then two, then any number of collisions, Paul Langevin succeeded in giving the solution to this problem. It was the last task he successfully completed before his death.

CURRICULUM VITAE

Born in Paris, XVIIIth Arrondissement, January 24, 1872.

Died in Paris, V0 arrondissement, December 19, 1946. Buried in the Pantheon, November 18, 1948.

1884. Student of l'École Lavoisier

1888. Student of l'École de Physique et de Chimie.

1893. Student of l'École Normale Supérieure.

1897. Agrégé of physical sciences.

1897. Fellow of the City of Paris at the Cavendish Laboratory, Cambridge. 1898. Fellow of l'École Normale at the Faculty of Sciences of Paris.

1900. Laboratory assistant the Faculty of Sciences in Paris.

1902. Replacement professor at the Collège de France.

1903. Substitute professor at the Collège de France.

1905. Professor at l'École de Physique et de Chimie.

1909. Full professor at the Collège de France.

1911 to 1927. Member of the first five Solvay Physics Conferences.

1920. Scientific director of the Journal de Physique.

1926. Director of l'École de Physique et de Chimie.

1928. Chairman of the Scientific Committee of the Solvay International Institute of Physics.

1930 to 1933. Chairman of the sixth and seventh Solvay Physics Conferences.

1934. Member of the Academy of Sciences of Paris.

1945. President of the Education Reform Commission.

Doctor Honoris causa from the Universities of Manchester, Leeds, Bristol, Cambridge, Brussels, Liège. Honorary professor at the University of Buenos Aires, honorary member of the Faculty of Sciences of Santiago de Chile.

Member of the Royal Society and the Royal Institution of London. Honorary Member of the Academy of Sciences of the U.S.S.R. Member of the Royal Society of Sciences of Gottingen, of the Academy of Lincei in Rome, of the Academy of Marine, of the Academies of Sciences of Prague, Bologna, Buenos-Aires, Copenhagen, of the Royal Academy of Ireland.

Grand-Croix de la Légion d'Honneur, Commander of the British Empire.

Note 1: A full bibliography of Langevin's 126 publications was appended to the obituary, dating from 1900 to 1950, which is omitted here. Nineteen of these are on topics associated with piezoelectricity and ultrasound.

Note 2: The translation was carried out in three stages. A first good draft was generated using Google Translate. This was then edited by the first author where needed to clarify the meaning, while retaining most of the original sentence structure. The final draft was proof-read by Philippe Blondel in order to ensure accuracy where this was uncertain.

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ANNEX

THE SCIENCE OF MEDICAL IMAGING
AN INTRODUCTION TO THE QUEST FOR VISIBILITY

P. Sprawls

Emory University School of Medicine, Atlanta, and Sprawls Educational Foundation. www.sprawls.org

I. INTRODUCTION

The ability to investigate and visualize the interior of the human body is the major method for diagnosing many diseases, injuries, and evaluating conditions in the practice of clinical medicine. It is also valuable for guiding and monitoring treatment and therapeutic procedures. It is the medical specialization known as Radiology, Roentgenology, or generally Medical Imaging. Radiologists are the physicians with the education, training, and Board certifications who use the variety of medical imaging procedures in clinical practice. Medical physicists are the predominant scientists in the field of medical imaging and radiology, with activities including research and development, clinical collaboration, and provision of education for all professionals working with medical imaging.

Physics is the primary science of medical imaging. The overall medical imaging process is illustrated in Figure 1.

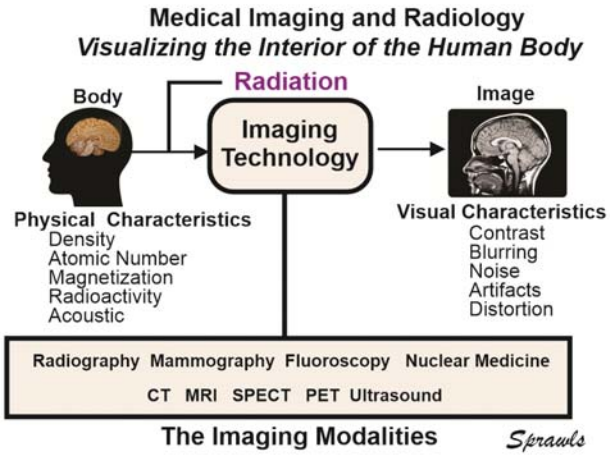

Figure 1. The elements of the medical imaging process.

The objective of a medical imaging procedure is to visualize some specific anatomical structure, object, or condition within the body. A medical image can be considered as a visual representation of some area of the human body, It is the process of converting physical characteristics of tissue within the body into visual characteristics within an image, as illustrated in Figure 1. The visibility of a specific item within the body depends on a combination of factors that must be considered in setting up and conducting an imaging procedure. The first is the selection of the modality, that is the type of equipment or technology to be used. The various modalities are identified in Figure 1. Each modality forms images using different physical principles and types of interactions with the tissue within the body. A distinguishing characteristic of a modality is the type of radiation used to penetrate and develop images through interactions with the tissue within a body.

The Imaging Modalities

Radiography, Mammography, and Fluoroscopy form images by projecting a beam of x-radiation through a body and producing "shadows" of internal structures and objects. The shadow images are formed by the varying attenuations of the x-radiation among the tissues that is determined by the physical characteristics of the tissue, especially physical density.

Nuclear Medicine includes a variety of procedures that uses radioactivity as the source of radiation. Radioactive substances are administered to patients and the gamma radiation from the body is imaged with a gamma camera. Nuclear Medicine using radioactivity also includes two of the tomographic modalities, SPECT and PET described below.

High-frequency, or Ultrasound, is a modality that has some specific clinical applications.

The differences among the modalities include the type of radiation used and the physical process used to form the images. Each modality has specific features, especially the characteristics of the images that determine visibility of specific conditions within the body, that make it "the modality of choice" for specific clinical applications. The choice of a modality for a specific clinical purpose--detecting breast cancer, for example, is the first decision a Radiologist makes. There are references for this, especially the Appropriateness Criteria published by the American College of Radiology (ACR).

Magnetic Resonance Imaging (MRI) is a tomographic imaging process that uses radio frequency (RF) signals to transmit information from the body tissues to the imaging system.

X-ray projection procedures (Radiography, Mammography, and Fluoroscopy) as well as the Gamma Camera, produce images of a body section. The chest is an example. All structures throughout the section produce shadows, with some overlying others. For many procedures this is not a problem and there is value in having one image that covers a complete anatomical region like the chest. However, there are advantages in having images of thin slices through a body section. This is *tomographic* imaging. The several tomographic modalities described below form images in two phases. The first phase is acquiring radiation data from a body, often referred to as "scanning." The second phase is mathematically "reconstructing" an image of slices within the body section from the data acquired during the scan..

The contemporary imaging modalities and methods available now are the result of over a century of research and development including the impact of the digital revolution. The results are imaging procedures with expanded *capabilities* for visualizing clinical conditions but also a major increase in *complexity*.

It is this *complexity* of the modern imaging procedures that requires a significant knowledge of the physics of the imaging process for the purpose of selecting imaging methods, evaluating image quality in relation to clinical requirements, and optimizing procedures with respect to image quality/visibility and potential risk to patients.

The Views

A distinguishing characteristic of a modality is the view of the human body it provides. There are two distinct views as illustrated in Figure 2.

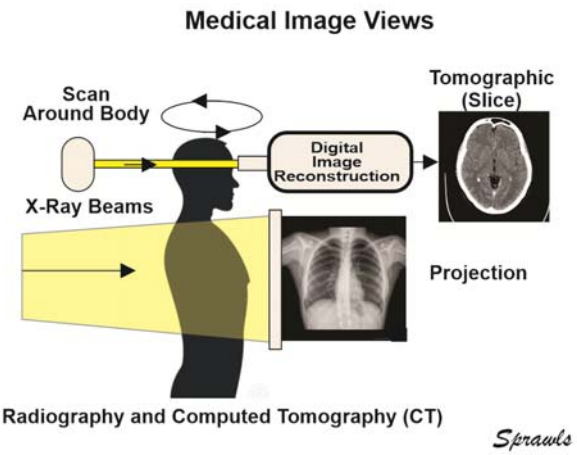

Figure 2. Comparing the two views of the human body, projection and tomographic.

II. Radiation

All medical imaging procedures require the use of some form of radiation or energy that can both penetrate the body and interact with the tissues within the body to form an image. This is the reason for the name, Radiology, or Roentgenology honoring Wilhelm Roentgen, the discoverer of x-radiation, who demonstrated its medical imaging capabilities. *X-radiation* was the first and continues to be the predominant form of radiation for medical imaging. *Gamma* radiation from radioactive substances administered to patients is used in the Nuclear Medicine procedures with the gamma camera, single photon emission tomography (SPECT), and positron emission tomography (PET). Both x-radiation and gamma are form of ionizing

radiation with potential biological effects (both good and harmful) to humans. The good is the use to treat cancer in the practice of Therapeutic Radiology or Radiation Oncology. There are potential harmful effects, generally related to the amount of radiation deposited in a body, that are considered in selecting and adjusting imaging procedures. Radiation Safety and Risk Management is one of the professional activities provided by physicists, generally designated as Health Physics.

Medical physicists are critical to this with their clinical activities focusing on maintaining image quality, risk management, as collaborators with radiologists and technologists, and as effective educators as required.

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III. The Big Picture

The medical imaging process is a complex system of elements that interact to produce images with specific characteristics that determine visibility. The overview or "big picture" shown in Fig. 3 provides a framework and foundation for an understanding of the imaging process and how to optimize it for specific clinical purposes.

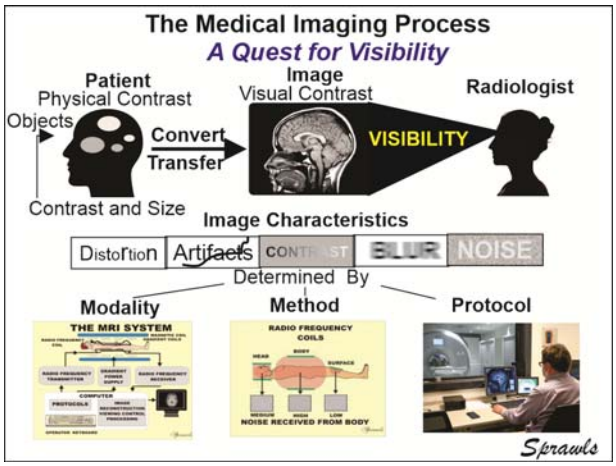

Figure 3. An overview of the medical imaging process and the many factors that determine visibility of specific conditions within the human body.

Most medical imaging procedures are a complex process with a combination of many variable factors that determine visibility of specific anatomical structures and clinical conditions. The radiologist is generally the professional with the responsibility for selecting and controlling the imaging procedure that is appropriate for a specific objective. This consists of first selecting a modality (MRI, CT, etc.) and an imaging method within the modality (Spin Echo, Inversion Recovery, etc.). The final step for each patient is the selection and adjustment of the *technique factors* (TR, TE, etc.) that form the *protocol* for the specific procedure. It is the combination of all these that determines the individual *physical characteristics* of images and the ultimate *visibility* of conditions within the body. It is the physical characteristics of the objects, structures, and conditions within the body that determine how they can be appropriately imaged (modality, method, and protocol).

IV. Physical Charactistics of Tissues and Fluids Within the Body

All medical imaging procedures are *physical interactions* between the imaging systems and the tissues and fluids within the body. The visibility of objects or conditions within a body depends on their physical difference, or contrast, relating to the surrounding area or background. Pathological conditions which are generally biological changes will be visible only if there are associated physical changes. For example, breast cancers are visible with mammography because they have an increased physical density in relation to the surrounding less dense tissue.

A major difference among the imaging modalities is the type of physical contrast that can be visualized.

X-Ray Imaging (Radiography, Mammography, and CT)

Radiography is the modality that produces recorded images by projecting an x-ray beam through the body and casting shadows of internal structures and objects. Typical radiographic images are shown in Figure 4.

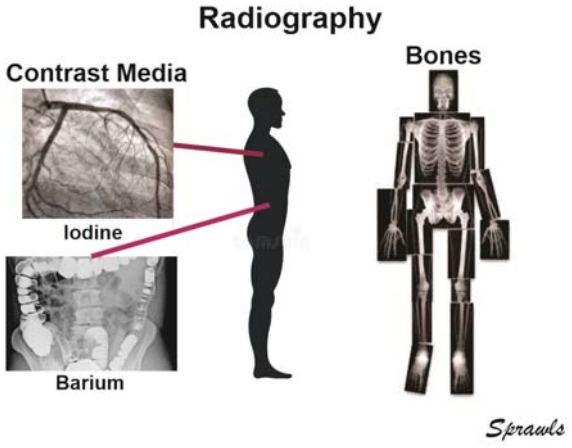

Figure 4. Radiographic images produced by natural contrast (bones) and administered contrast media (iodine and barium).

Radiography, including mammography, and fluoroscopy are projection imaging procedures in which an x-ray beam is passed through the body and casts shadows of internal anatomical structures and objects. All x-ray images, including CT, are formed by the difference in x-ray absorption/attenuation among the tissues. The two physical characteristics of tissue that determine x-ray attenuation are physical density and atomic number (Z). The effective atomic number (Z) of tissues, approximately 7.xx, has relatively little variation among the soft tissues and fluids in the body and is not a major source of contrast. It does contribute to the contrast and visibility of calcium and bones. The natural and often limited contrast among the soft tissue, fluids, and organs can be enhanced by administering substances, known as contrast media, during an imaging procedure as illustrated in Figure 4. The atomic numbers of iodine and barium enhance their x-ray attenuation and contribute to their effectiveness as contrast media. Compounds containing iodine are used for enhancing the visibility of blood vessels and the urinary tract. Barium is used for imaging the digestive tract including stomach and intestines. Differences in physical density are sources of contrast especially in the chest with the low-density air in the lungs providing a background for the bones, heart, and more dense lesions and cancers. Mammography is a special form of radiography for imaging the breast. It uses special types of x-ray beams to image the small differences in density within the breast, especially between the normal tissues and cancers.

Computed tomography (CT) is an x-ray imaging modality that has the capability to produce visible contrast among the small differences in density of the soft tissues, as in the brain. As illustrated in Figure 2, CT is a tomographic imaging procedure in which a thin x-ray beam is scanned around the patient in the plane of the slice of tissue being imaged. The beam measures the x-ray attenuation along many pathways or projections through the tissue slice. A computational process known as *image reconstruction* uses the attenuation data to produce an image that is a display of density differences among the tissues. The two great values of CT are the production of tomographic/slice images without interference from overlying anatomical structures and its high sensitivity for visualizing small differences among tissues.

Magnetic Resonance Imaging

Magnetic Resonance Imaging (MRI) is a tomographic imaging process somewhat like CT but different in the physical tissue characteristics that are visualized. An MR image is an image of *magnetized* tissue. During the imaging procedure the tissues and fluids in the body are temporarily magnetized by a strong magnetic field. The level of magnetization of a specific tissue is determined by its hydrogen content, the only significant chemical element in the body with magnetic nuclei (protons) that align to produce the tissue magnetization. The concentration of hydrogen, or proton density (PD) is a characteristic of each specific tissue and a source of contrast. During the imaging process the magnetic nuclei are periodically flipped from their stable or relaxed direction. This is followed by a period of realigning or relaxation in relation to two different magnetic field

directions, longitudinal and transverse. The times required for these relaxations, designated as T1 (longitudinal) and T2 (transverse), are characteristics of each specific tissue. This provides visible contrast among both the normal tissues, like gray and white matter in the brain, and between normal tissue and many pathological conditions…one of the great values of MRI.

Each MR image is an image of the *level of magnetization* in each tissue. However, the level of magnetization is constantly changing as the tissues go through the relaxation cycles. Therefore, the level of magnetization, or brightness of a tissue displayed in an image depends on "when the picture was snapped" during the relaxation cycle. This is determined by adjustment of factors in the imaging protocol. Generally, the imaging protocol can be set to produce images where each of the tissue characteristics, PR, T1, and T2 are the major source of visible contrast. It is the combination of the tissue characteristics and image protocol factors that determine the level of magnetization of each specific tissue and the resulting brightness in am image. Figure 5. compares MR and CT images and their sources of physical contrast.

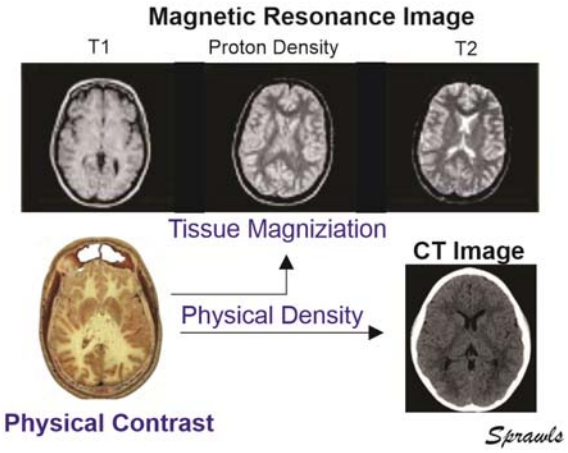

Figure 5. Two sources of physical contrast, density, and magnetization, that contribute to visibility of tissues.

Medical imaging is a physical process and can only produce images of the physical contrast within the body. There is naturally occurring physical contrast that provides for some visualization of most anatomical regions and organ systems of the body. When that is not adequate different forms of contrast media can be administered to enhance visibility of structures (GI track, blood vessels, etc.) and pathological tissues.

Radionuclide Imaging (Nuclear Medicine)

A form of *physical contrast* can be created in the human body with the administration of a radioactive substance that selectively concentrates in different tissues relating to conditions, especially pathologic, and function such as blood perfusion in the myocardium or metabolic activity in the brain. A variety of radionuclides are used. Images displaying the distribution of radioactivity in an anatomical region or organ are produced with three specific modalities.

The *gamma camera*, as the name implies "takes a picture" providing visibility of radioactivity in the body.

Single Photon Emission Tomography (SPECT) uses a gamma camera that is rotated around the patient's body and uses image reconstruction to produce tomographic images. It is a form of computed tomography using radioactivity as the contrast to be imaged. Positron Emission Tomography (PET) is also a tomographic method that provides visibility of radioactive nuclides that emit positrons that can be used to image metabolic activity. These three modalities are compared in Figure 6.

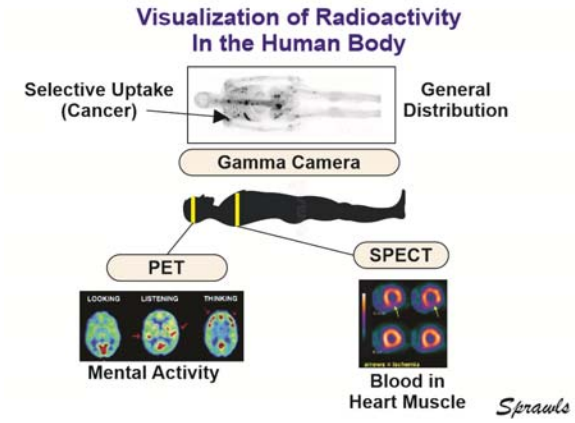

Figure 6. The three imaging modalities for visualizing radioactive pharmaceuticals that have been administered to a patient.
A major distinction is that the gamma camera views a large section of the body, like a photographic camera, and produces an image of the radioactivity through that section. Both PET and SPECT are tomographic imaging methods producing images of selected slices.

Ultrasound Imaging

High-frequency sound, ultrasound, can be used to produce images within the human body with two types of interactions. One is by reflections from structures within the body that produce echoes; the other involves Doppler shifts in the frequency of sound produced by the motion of flowing blood. These are illustrated in Figure 7.

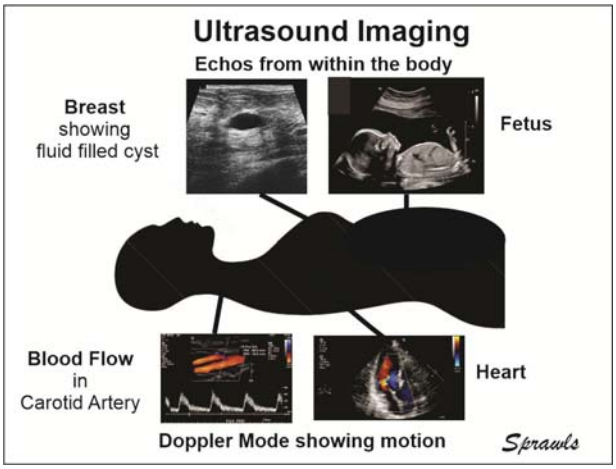

Figure 7. Images produced with ultrasound with the two modes: echoes from reflecting structures and visualization of flowing blood using the Doppler effect.

An ultrasound image is a display of structures or objects within the body that reflect the sound pulses transmitted into the body. Reflections occur at surfaces or boundaries between tissues that have different acoustical properties, especially the speed of sound. Most organs and tissues within the body are a mixture with different characteristics and produce echoes that are visible in an image. An exception is a clear fluid like water that does not produce echoes. This is illustrated in the image of the breast where the fluid-filled cyst does not produce echoes and appears as a "black hole". This is especially valuable for distinguishing fluid filled cysts from tissue masses that could be cancer. Ultrasound is the preferred method for imaging fetuses before birth. It is a quick and easy procedure as it does not use ionizing radiation, x-rays, with a potential risk.

V. Digital images and image structure

Virtually all medical imaging modalities produce images in a digital format, very different from images on film as in the past. There are many values and advantages with digital images, like digital photography that we all use, easy to adjust, store, and send to others. However, the structure and numerical dimensions of digital images have a major effect on the several image quality characteristics and visibility. The digital image *numerical dimensions* are often some of the procedure protocol factors that can be adjusted to optimize the image quality and visibility for a specific patient procedure.

There are two distinct functions that occur in the production of digital medical images with all the modalities. One function is creating a numerical value for the tissue characteristic that is being imaged and the other is the formation of an image displaying these numerical values over the viewed section of the body. The difference among the modalities is how these functions are performed. Here we consider the common characteristics that apply to all modalities. The first, developing numerical values for the tissue characteristics, is illustrated in Figure 8.

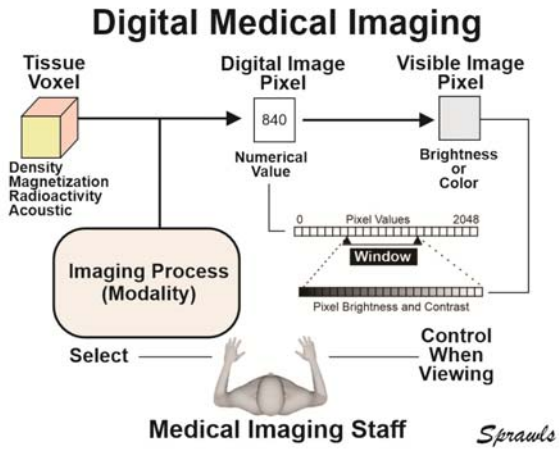

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Figure 8. The two phases of creating a visible image displaying the physical characteristics of tissue within the body.

The first phase is the acquisition of the data from the body, often referred to as "scanning" and the creation of numerical values, a digital image. The common function of all imaging modalities is that the body is divided into small cubes of tissue, voxels, and the physical characteristic of the tissue is measured, with the value recorded in a corresponding area in the image, a pixel. Which tissue characteristic is being imaged is determined by the modality selected by the medical staff. The second phase that can take place later is the viewing of the digital image by the radiologist or other medical professional. When viewing a digital image, the conversion of the digital image into a visible image can be controlled to emphasize the visibility of specific objects and areas within the body. The primary control is the "window" that selects the range of digital values that will cover the brightness range in the visible image.

The other function within an imaging procedure is the formation of the tissue voxels into a matrix, typically representing a slice of tissue within the body, and a corresponding image as a matrix of pixels as illustrated in Figure 9.

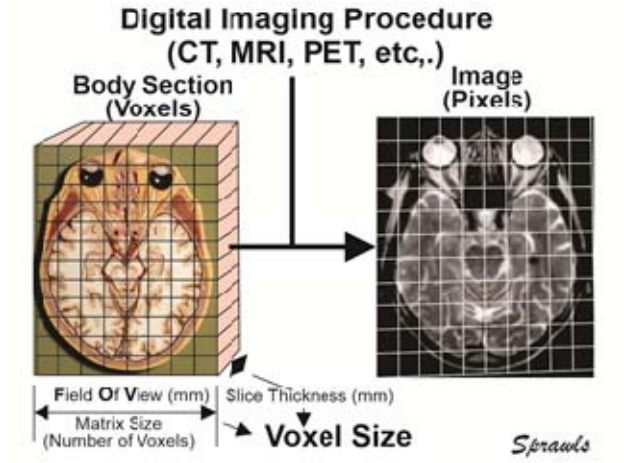


Figure 9. The dimensions associated with a digital image that affect image characteristics and visibility.

All the medical imaging procedures that produce digital images begin by dividing the area of the human body that is to be imaged into a matrix of small cubes known as "volume elements" (voxels). The image is formed as a matrix of "picture elements" (pixels). The size of the voxels is a major factor determining the quality characteristics of the image, especially the blurring and visual noise.

Each voxel can be considered as a discrete "sample" of tissue. In general, the imaging process measures a physical characteristic of the tissue (density, magnetization, or radioactivity) and calculates a numerical value. The numerical value is then displayed in a corresponding image pixel as a brightness. For example, in a CT image the brightness of each image pixel is determined by the physical density of the tissue in the corresponding voxel. In MRI the brightness of a pixel is determined by the magnetization of the tissue in the corresponding voxel.

Our interest here is the effect of voxel size on the quality characteristics of the image. As illustrated, the size of a voxel is determined by three (3) dimensions as shown. These dimensions are established for each of the imaging modalities in relation to the physical characteristics of the imaging equipment and the imaging process. Generally, with each modality, the dimensions, especially slice thickness, can be adjusted or "fine-tuned" when setting up the procedure protocol for a specific patient.

The significance of the dimensions of the digital images, the voxels, and pixels, is that they are major factors determining the quality characteristics and visibility with specific images. They must be selected and adjusted to provide appropriate visibility with each medical imaging procedure. It is not a simple process because of the conflicting effects on image characteristics and visibility that some of the dimensions have and the necessity to develop *optimized imaging protocols* for specific patient procedures, an application of physics to clinical medicine.

VI. IMAGE CHARACTERISTICS THAT DETERMINE VISIBILITY

The visibility of specific anatomical structures, objects, or conditions within the human body is determined by a combination of image characteristics as illustrated in Figure 10. These characteristics and the resulting visibility are determined by a complex combination of factors including which *modality* is selected, the imaging *method* within the modality, and the adjustment of the specific imaging *protocol* with respect to the *technical factors* adjusted for the procedure. The objective here is to consider the individual image characteristics and their effect on visibility. This is the fundamental science of medical imaging.

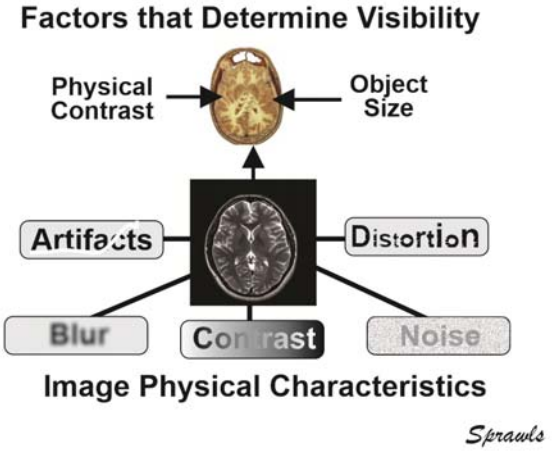


Figure 10. The combination of image characteristics that determine visibility of objects within the body.

Visibility is a goal of all imaging procedures, ranging from normal human vision to the use of satellites orbiting the earth. Many forms of technology--microscopes, telescopes, television, etc.--are used to extend the range of human visibility to see objects that are invisible because of conditions including size, distance, enclosures, and other characteristics and composition of the objects. Medical Imaging Systems are the technology used to extend human vision into the human body. As described before there are a variety of technological systems, the *modalities*, used in modern medical imaging procedures. As illustrated in Figure X, it is the choice of modality, the imaging method within the modality, and the adjustment of protocol technical factors for each patient procedure that determine the physical characteristics of an image and visibility.

There are five (5) fundamental characteristics of images as illustrated in Figure 10 that collectively determine visibility of specific objects within the body. It is a complex process because of potential conflicting relationships among the characteristics and dependence on factors including limitations of the technology, radiation exposure to patients, and the time required to produce images.

The fundamental science of medical imaging, physics, is built on an understanding of the physical characteristics of images and their relationship to visibility.

VII. Image Contrast and Procedure Contrast Sensitivity

Contrast means *difference*, the difference between and among items, and is the fundamental characteristic of an image. Within an image an object is visible only if it is *visually different* with respect to the surrounding area or background, either in brightness or color. Without contrast there is no image. The contrast contributing to visibility originates as physical contrast within the body as physical differences among tissues or objects to be visualized as described previously. The imaging process converts the physical contrast and transfers it to an image as visible contrast as illustrated in Figure 11.

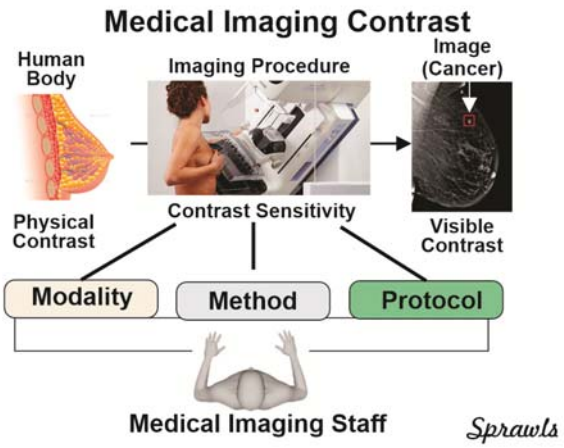


Figure 11. The factors that determine and control the visible contrast in a medical image.

The objective is not to produce as much image contrast as possible; it is to produce sufficient contrast for the desired visibility. The transfer of *physical contrast* in the body to *visible contrast* in the image is determined by the *contrast sensitivity* of the imaging procedure. Contrast sensitivity is the major characteristic of an imaging procedure for visualizing specific objects or conditions within the body. A cancer in the breast is the example used here. The contrast sensitivity is determined first by the selection of an appropriate imaging modality and method, and then adjusting the technical protocol factors for the specific patient.

The contrast sensitivity of an imaging procedure determines the lowest physical contrast within the body that will be visible as illustrated in Figure 12.

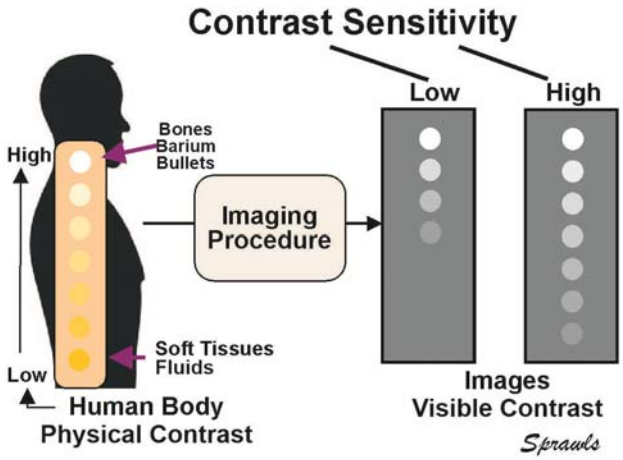


Figure 12. The concept of contrast sensitivity: the characteristic of an imaging procedure that determines the visibility of objects with the lowest physical contrast within the body.

It is not difficult to produce visible images of high-contrast objects in the body, such as bones, bullets, etc. The continuing challenge is to develop procedures to visualize the low contrast objects, especially the soft tissues and fluids. Mammography is the modality that has developed over the years with innovations to enhance contrast sensitivity and the visibility of cancers. A major innovation in medical imaging was the invention and development of computed tomography (CT) that provided visualization of the soft-tissue brain within the dense skull as illustrated in Figure 13.

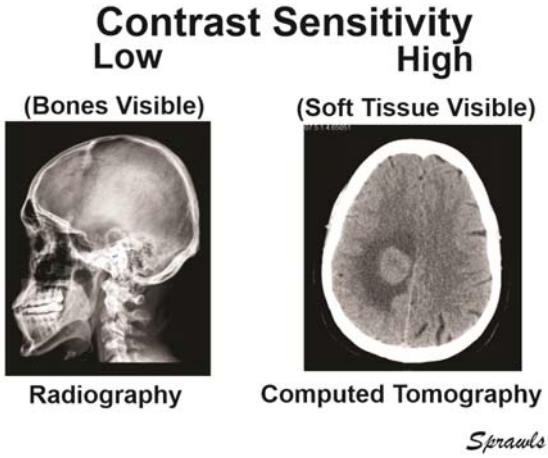


Figure 13. Comparing the contrast sensitivity of two x-ray imaging modalities, radiography and computed tomography (CT).

Radiography produces images by projecting an x-ray beam through the body and casting shadows of objects with high physical contrast, especially the bones. As illustrated in the image above low-contrast soft tissue brain is not visible in the radiograph. However, with CT, which is also an x-ray imaging process, the low physical contrast among the soft tissues, and especially diseased tissues like the tumor, is visible. This high *contrast sensitivity* is the major characteristic of CT that contributes to its value for viewing soft-tissue organs throughout the body. The combination of two factors contributes to this. It is a tomographic process as illustrated in Figure 2 that displays images as thin slices without looking through bones and it uses digital image processing as illustrated in Figure 14.

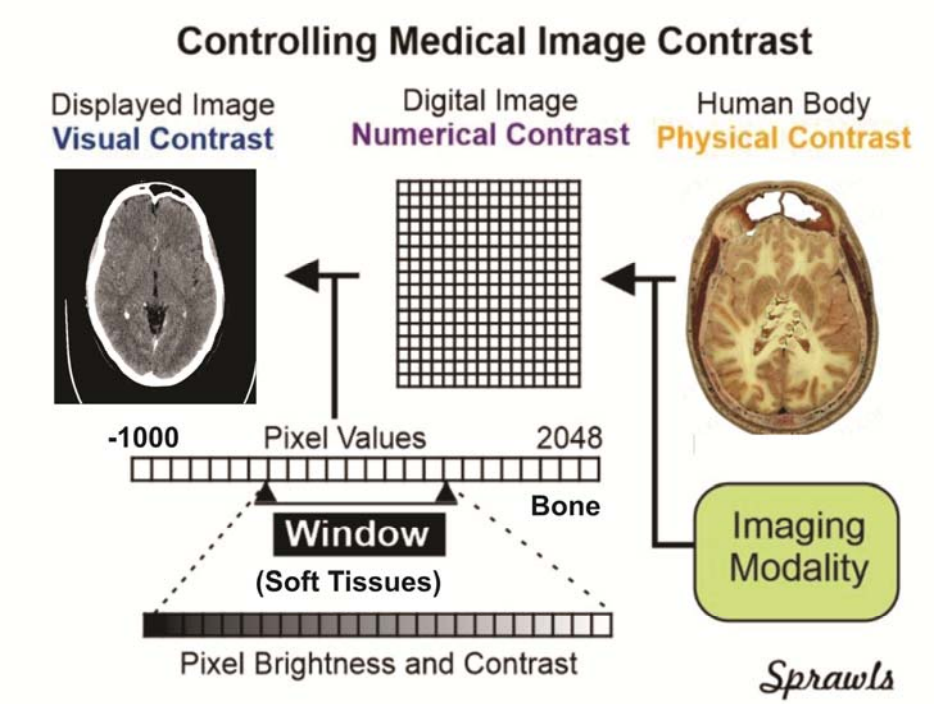


Figure 14. Using digital image processing to provide high contrast sensitivity and control image contrast to enhance visibility.

Computed tomography first creates a digital image by the process of mathematical image reconstruction from data collected on the attenuation of x-ray beams passing through slices of the human body. The numerical value of each pixel is determined by the physical density of the tissue in the corresponding voxels. The pixel values cover a wide range from low-density air (-1000) to high-density bone (2048). The soft tissues and fluids cover a small range within the large range. As shown, when a digital image is being displayed and viewed, the range of pixel values that will cover the brightness range (from white to black) is adjusted with "window" control. With this, the relatively small range of soft tissue densities and physical contrast can be displayed over a large brightness range and visible contrast.

The ability to control the window when viewing medial digital images provides the opportunity to adjust and optimize the contrast and visibility of various anatomical objects and regions within an image.

Medical physicists evaluate the contrast sensitivity of medical imaging procedures with test devices/phantoms that contain a series of objects with varying physical contrast and determining the lowest physical contrast that is visible. One of these will be illustrated later.

As described and illustrated previously each of the modalities visualizes different forms of physical contrast, tissue density, tissue magnetization, radioactivity, etc. The contrast sensitivity of the imaging process determines the lowest level of physical contrast that can be visualized. This is especially significant because the physical contrast, especially among the soft tissues which have almost similar density values, is often very low. Mammography is a special type of radiography that has been developed to have a high contrast sensitivity. Within a *modality* there are typically a selection of *methods* that can be used. With modern mammography there is the choice of views, and either projection or tomographic imaging which affects contrast sensitivity. Then for each patient there are various technical, or technique factors that are adjusted often depending on the size of the patient or specific clinical needs. The selection and adjustment of these factors establish the procedure *protocol* for each patient. The number of factors in a protocol for a specific patient can range from just a few for radiography to perhaps 20 or more for the much more complex imaging modalities like MRI.

Medical physicists provide the scientific and technical knowledge and experience needed to ensure adequate contrast sensitivity and visibility for imaging procedures. This is through research and development of imaging technology and methods, evaluation of equipment performance (quality control and assurance), collaboration with physicians and imaging staff attempting to optimize procedure protocols, and especially to provide educational opportunities for the other medical imaging professionals. In these activities physicists use a variety of test objects, often referred to as phantoms, to measure and evaluate contrast sensitivity. This will be illustrated later.

VIII. Blurring and visibility of detail

With all forms of imaging, including human vision, the size of objects affects and limits their visibility. This is generally designated as *visibility of detail*. It is the *blurring* that occurs during an imaging process that limits *visibility of detail*. The reality is there is some amount of blurring in all imaging procedures, including human vision. It is the blurring within our visual system, that often increases with age, that limits the visibility of small print and other small objects--that is, visibility of detail. This characteristic, visibility of detail (the medical term is visual acuity), is tested with a chart shown in Figure 15.

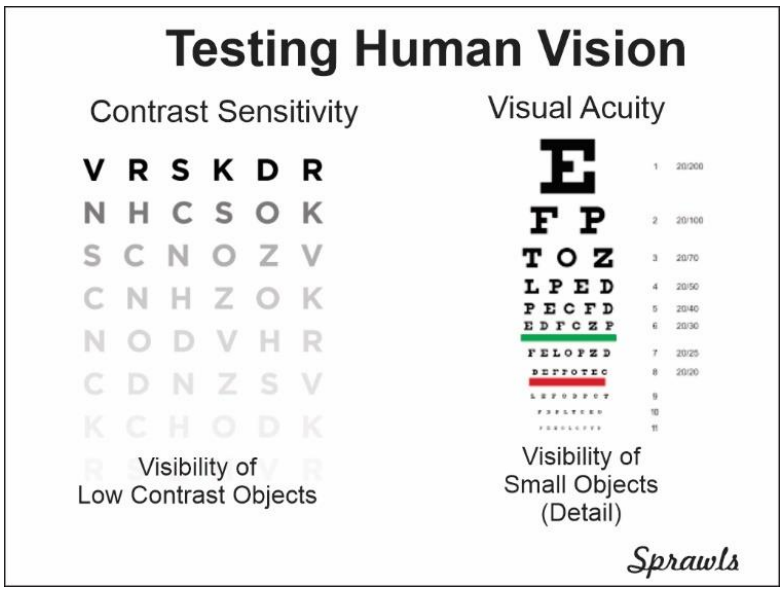


Figure 15. A chart used to test the two factors, contrast sensitivity and blurring, that limit visibility in relationship to the objects being viewed, their contrast and size.

Blurring, which is present in all imaging systems, from the human eyeball to satellites circulating and viewing the earth, limits visibility of objects in relationship to their size, often referred to as image detail. In optical systems cameras, projectors, etc., blurring is generally caused by inadequate focusing of the lens. There is some blurring that occurs in all medical imaging procedures and is the factor that limits the visibility of small objects and structures, or detail in the body. This is a significant factor because it determines which objects and structures can be viewed and the clinical procedures that can be performed with each of the imaging modalities. An example, mammography, is illustrated in Figure 16.

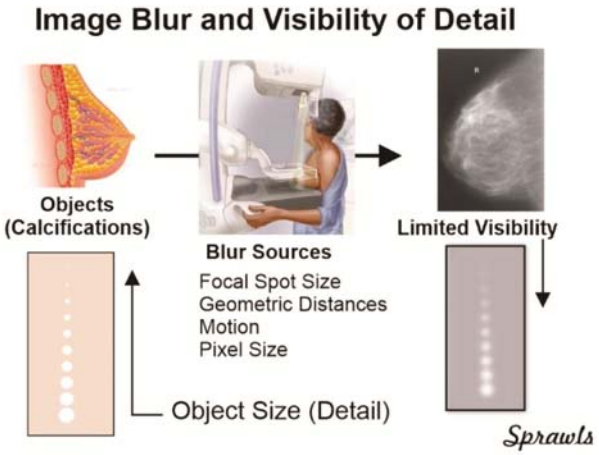


Figure 16. Sources of blur and effect on visibility of small objects and detail within the body.

The blurring with each imaging modality is determined by the physical design of the equipment and the process for producing the images. In mammography, where the image is produced by projecting an x-ray beam through the breast, the size of the x-ray source (the x-ray tube focal spot) and the distances within the setup are sources of blur along with any motion of the patient during the formation of the image. Of all the imaging modalities, mammography is designed to produce images with the least amount of blurring. This is because a valuable sign of breast cancers are small, micro-calcifications (approximately 0.15 mm) that need to be viewed for a diagnosis,

A significant source of blurring with all imaging modalities that produce digital images is the size of the tissue voxels and image pixels. Voxel and pixel sizes are design characteristics of each modality and can often be adjusted within each modality through three protocol technique factors: anatomical field of view, matrix numerical size, and slice thickness for the tomographic imaging procedures. This is illustrated for mammography in Figure 17.

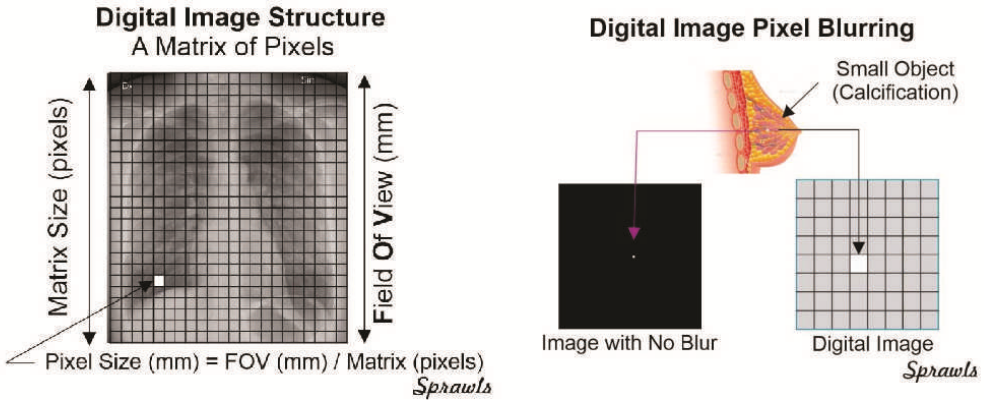


Figure 17. Factors that control pixel size and the blurring produced by pixels in a digital image.

A tissue voxel or image pixel is a blur because all objects or structures within it are mixed or blurred together and represented by one numerical number.

A defining characteristic of each medical imaging modality is its inherent blurring and visibility of detail it can provide as compared in Figure 18.

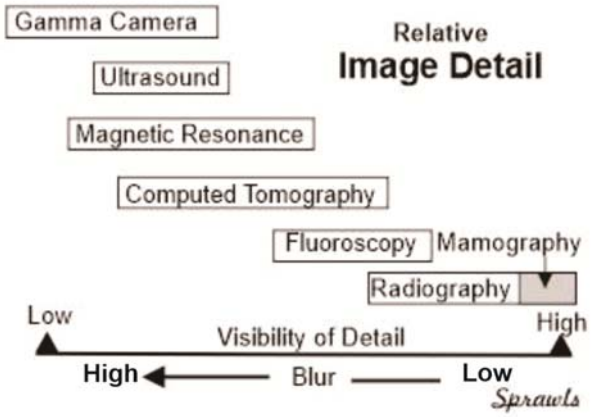


Figure 18. The relative visibility of detail provided by the imaging modalities.

The general visibility of detail (amount of blurring) with each modality is determined by the design of the equipment and how the images are produced. For each modality, computed tomography (CT) for example, there is a range of blur values and visibility of detail that can be provided. There is the selection of specific *methods* within a *modality* and especially the adjustment of the *technique factors* to establish the imaging protocol for each patient. The visibility of detail for a specific image is determined by the combination, or composite of the several sources of blur within the imaging procedure. In principle, the blurring and resulting visibility of detail can be adjusted over a relatively wide range for each imaging modality is illustrated in Figure 19.

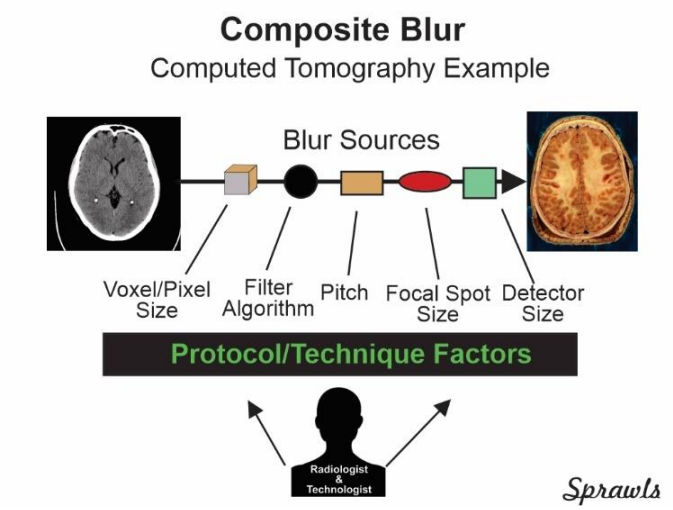


Figure 19. An example of the factors that can be used to adjust the blurring and visibility of detail in a specific imaging procedure.

That raises a question. If the blur is adjustable, why not set it to the lowest value so that images provide maximum visibility of detail? As described later, many factors that affect blurring also affect other image characteristics and the radiation exposure to patients. These are often opposing factors. Changing a factor to *reduce blurring* can *increase visual noise* resulting in less overall visibility and image quality.

The imaging protocol for a specific patient procedure should be *optimized* to provide an appropriate balance among the image characteristics and other factors including radiation exposure to the patient. Medical physicists provide the knowledge and experience to support this process.

IX. Visual noise in images

We are most familiar with audio noise--sounds that interfere with our hearing and often are distracting and perhaps irritating. There are other types of noise in other forms of communication including electronic signals. Our interest here is *visual noise* that interferes with *visibility* in medical images. Visual noise is an image of *random objects*, specs, or spots, superimposed over the image that is to be viewed. It is sometimes known by other names, including grainy or mottle. In medical imaging there are two major sources of visual noise. In the modalities using x-radiation or radioactive materials (so-called ionizing radiation) it is the random nature of photon production and interactions. In MRI that uses radiofrequency (RF) signals to acquire images from the human body, noise is produced by stray RF energy created by thermal activity with the body tissue.

The specific effect of visual noise is to *reduce visibility* of low-contrast objects within an image as illustrated in Figure 20.

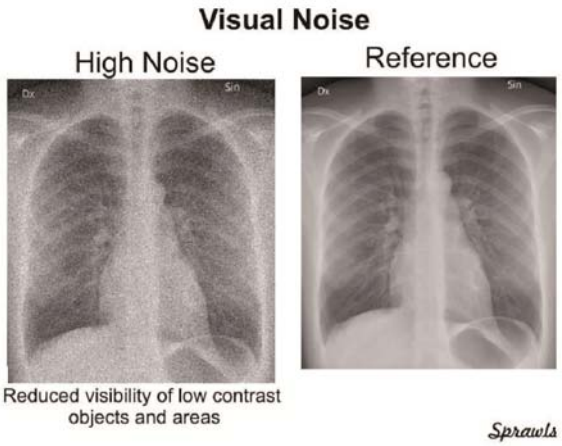


Figure 20. The effect of noise on visibility in an x-ray image.

Visual noise is an image characteristic that limits visibility of certain objects or structures within the body. Specifically, it reduces and limits visibility of objects that have low visual contrast. This is typically within the soft tissues and organs within the body and not with imaging the high-contrast bones. Some level of noise is in all x-ray images but can be controlled. The noise is an image of the x-ray beam itself that in added to the image of the patient anatomy as illustrated in Figure 21.

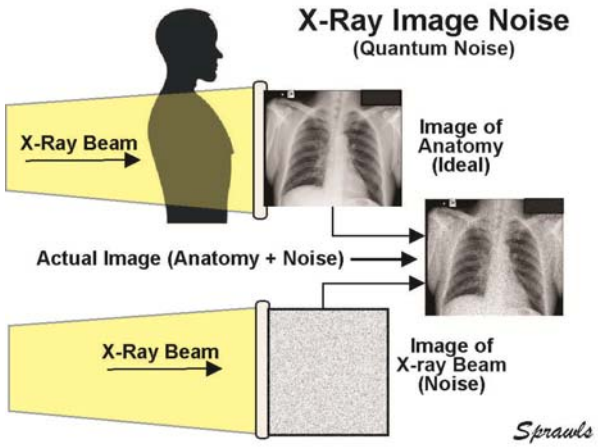


Figure 21. The source of noise in an x-ray image.

The control of the noise in an x-ray image is by adjusting the intensity (exposure) of the x-ray beam which determines the natural random distribution of the x-ray photons within the beam as illustrated in Figure 22.

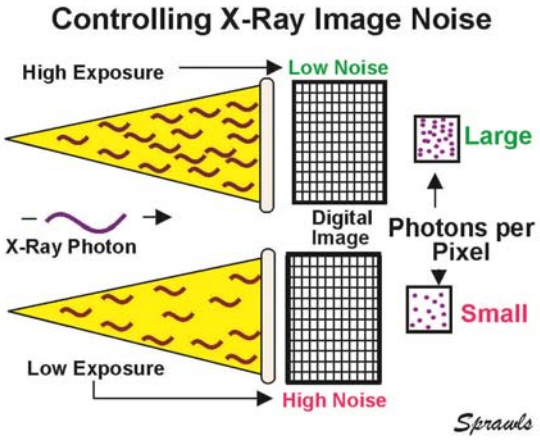


Figure 22. The relationship of x-ray image noise to the intensity of x-ray beam (exposure) forming the image.

This is perhaps the most significant factor in the physics of medical imaging because it affects two of the image characteristics that control visibility (blurring and noise) and the amount of radiation (dose) deposited in a patient's body. It is the factor that requires physics knowledge and experience in *optimizing imaging procedures* for maximum benefit for individual patients.

X-radiation and radiation from radioactive materials used for imaging is in the form of small units (quanta) of energy, photons. An x-ray beam can be considered as a "shadow" of individual photons, as rain is a shadow of individual drops of water. The significance is they are randomly distributed with respect to both area and time. Consider the example in Figure 16 where a digital image is formed by exposure with a "uniform" x-ray beam. The reality is at the atomic scale level an x-ray beam is not uniform but has a non-uniform and random distribution of the photons. In the formation of a digital image this results in a random distribution of the *number of photons* captured in each pixel. This produces a random variation in the brightness of the displayed pixels in the image which we see as visual noise as shown in figure 23.

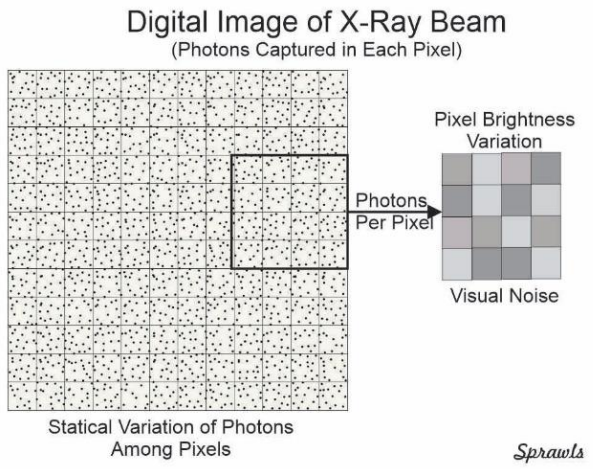


Figure 23. The source of visual noise in an x-ray image.

This statistical variation which appears as noise is *inversely related* to the average number of photons captured in each pixel, or in each voxel for tomographic imaging procedures. With knowledge of the statistics, it is possible to control the noise in images and set it to a level that is appropriate when considering other factors, especially radiation exposure to patients. The statistical distribution is shown in Figure 24.

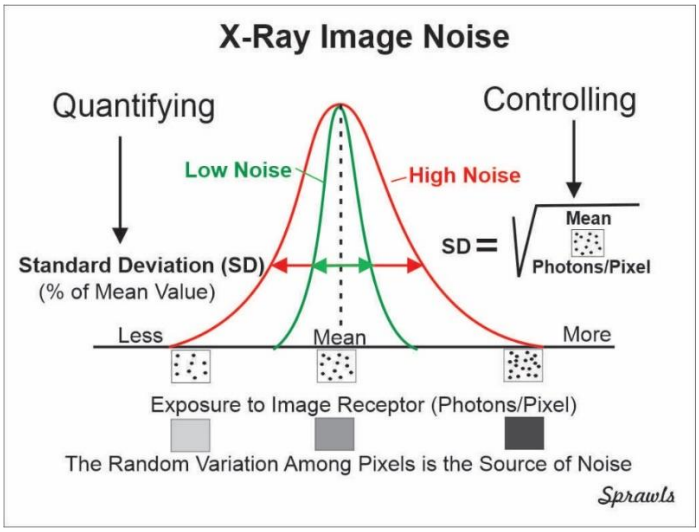


Figure 24. The relationship of the random variation of photons among pixels (the source of noise) and the average number of photons (exposure) in each pixel.

The variation of photons among the pixels follows a statistical *Gaussian distribution* with a specific property. The variation that is the source of the noise is inversely related to the number of photons forming each pixel in the image. This makes it possible to both quantify or calculate the noise and control it in an image.

In a Gaussian distribution the spread or distribution of photons can be calculated and expressed as the statistical quantity, Standard Deviation (SD), The value of the SD is the mathematical square root of the average number of photons per voxel as indicated in Figure 24.

The significance of this is that the noise in x-ray and radionuclide images can be controlled by adjusting the exposure to the imaging system which also affects the exposure to patients. A major decision that must be made in selecting imaging modalities and methods along with adjusting the technical factors in the protocol for specific patients is to achieve an appropriate balance between *image visibility* and *radiation exposure* to patients. For the modalities that produce images in a digital format, which is the majority that are used now, it becomes a complex issue because image blurring becomes a factor. This is because of opposing effects of voxel/pixel size on both visibility of detail (blurring) and visual noise as illustrated in Figure 25.

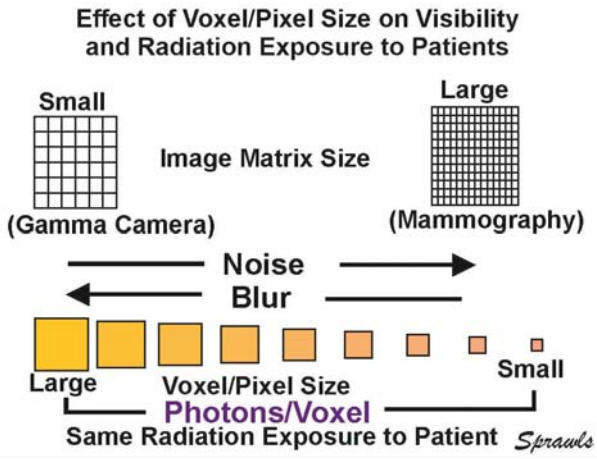


Figure 25. The opposing effect of voxel/pixel size on two factors determining visibility, blur and noise

The size of tissue voxels and corresponding image pixels varies over a large range for the imaging modalities. The factor that accounts for most of this variation is the numerical size of the image matrix. This is first determined by the design of the imaging equipment for each modality and can then be adjusted some by the imaging staff when setting up the procedure protocol for a specific patient. The voxel size directly affects the image characteristics as illustrated here. It indirectly affects another major factor--the radiation delivered to the patient's body.
When voxel size is reduced to reduce blur and improve visibility of detail the noise will be increased because the smaller voxel will capture less photons. This will require an increase in the radiation exposure to reduce the noise to an acceptable level. These are the three factors that are taken into consideration in the design of optimized procedure protocols for patients. This applies to all imaging modalities using x-radiation and radiation from radionuclides.

Visual noise is a significant factor affecting visibility with magnetic resonance imaging (MRI) but the source is different from the other imaging modalities. With MRI radiofrequency (RF), signals transmit the tissue characteristics from the body to produce an image. Unfortunately, the thermal activity within the human body generates some random RF energy that appears as noise in images as illustrated in Figure 26.

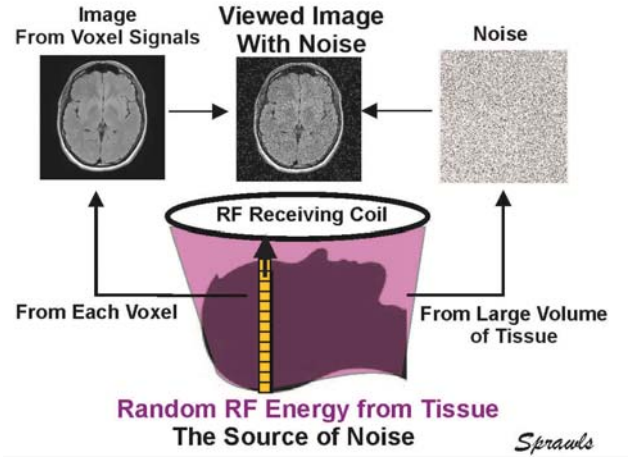


Figure 26. The source of visual noise in MRI.

Magnetic Resonance images are produced by receiving and processing radiofrequency (RF) signals from each tissue voxel that provides a measure of the tissue magnetization at that specific time in the imaging process. These signals are stimulated (and are like echoes) by RF pulses transmitted into the body during the imaging procedure. The intensity of the signal from each voxel determines the brightness of the corresponding image pixel. Unfortunately, some low-level RF energy is generated by the thermal activity within the human body. This random RF energy is also received by the imaging system and appears in the image as noise as illustrated in Figure 26.
The quality of the image and visibility of the tissues is determined by the ratio of the *signal to noise* strengths. The visibility of the noise is reduced by increasing the *signal to noise* ratio. Actions to achieve this including larger voxels, stronger magnetic fields, the special design of the RF receiving coils, and the collection and averaging of a larger quantity of RF signals with extended imaging procedure times. Some of these (magnetic field strength) are determined by the design of the equipment but most are factors that can be changed and adjusted when setting up the procedure protocols for specific patients.

X. Image Artifacts and Distortion

Contrast, blurring, and visual noise are characteristics that apply to *all medical images* and collectively determine visibility of the objects and structures within the body that are important for medical diagnosis. There are two other image characteristics, artifacts and geometric distortion, that are not in all images but are generally undesirable. When artifacts are present, they generally do not reduce overall visibility within the image but are distracting and might cover some objects or areas. Some examples are shown in Figure 27.

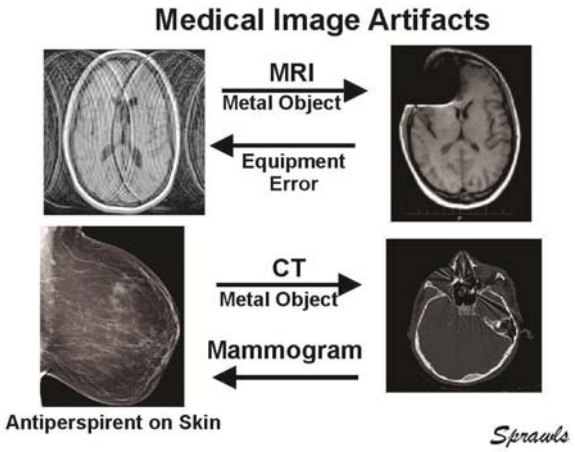


Figure 27. Some example artifacts that appear in medical images.

Artifacts are features that appear in an image that are not caused by the body anatomy itself. Artifacts are sometimes referred to as "ghosts" and not representing actual objects or structures. In general, the types and sources of artifacts are specific to each of the imaging modalities relating to how the images are produced. With radiography and mammography where an x-ray beam is projected through the body, non-anatomical objects in the path of the beam, both internal and on the surface, produce shadows that appear in the image as artifacts. While many will be recognized and identified for what they are, some can be mistaken and incorrectly diagnosed for signs of a disease. An example in Figure 21 are small pieces of an antiperspirant cosmetic, On the surface of the body they can appear as abnormal and medically significant objects within the breast. Implanted metal objects within the body are common sources of artifacts in both MRI and CT because they interfere with the imaging process, usually appearing as white streaks or dark areas. For both MRI and CT where data to create the image is acquired over an extended time, movement by the patient, including breathing, during this time distorts the data and can produce artifacts in the image. Within the imaging modalities there are a variety of techniques and actions that can be taken to reduce specific artifacts.
Distortion is a characteristic of an image in which the size and location of objects and structures are not accurately displayed. The most common is in radiography where there is geometric magnification of objects at different depths within the body which will be magnified differently.

XI. Evaluating Image Characteristics and Visibility

As described above visibility in a medical image varies over a large range and can be controlled by the selection of imaging modalities, methods within a modality, and the adjustment of technique factors for the procedure protocol for specific patients. To provide appropriate visibility with imaging procedures it is necessary to evaluate or measure visibility and the effects of the variable image characteristics, contrast, blurring, and noise.

Radiologists generally do this based on their experience in viewing many images and their judgment on image characteristics appropriate for specific clinical procedures.

Medical physicists use a more scientific method by producing and analyzing images of test devices or phantoms. One such device is illustrated in Figure 28.

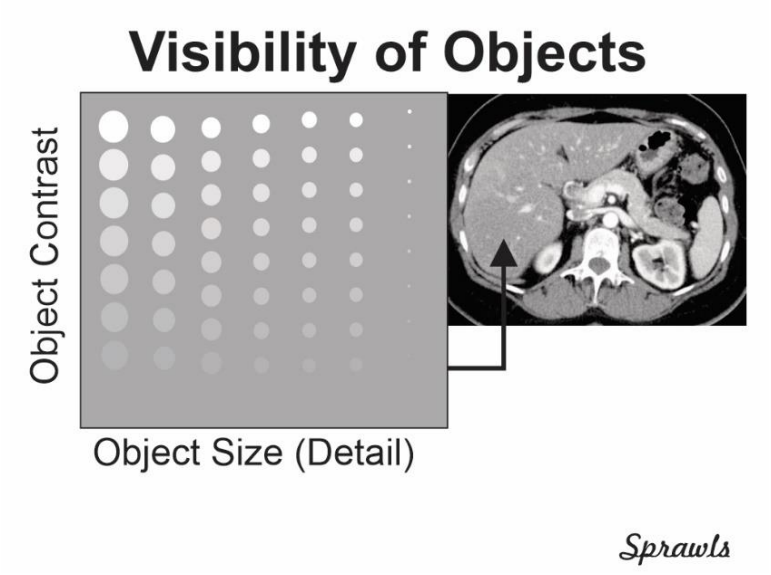


Figure 28. A Contrast Detail test device or phantom (shown on the left) contains objects with varying contrast and size/detail to represent the range of objects in the human body.

Physicists evaluate a specific imaging procedure by imaging the phantom and then analyzing the image to determine which objects are visible. This is a test of both contrast sensitivity and blurring. With reduced contrast sensitivity the objects with lower contrast will not be visible. Blurring reduces the visibility of the smaller objects.

A Contrast Detail phantom can also be used to evaluate the effect of noise on visibility as illustrated ion Figure 29.

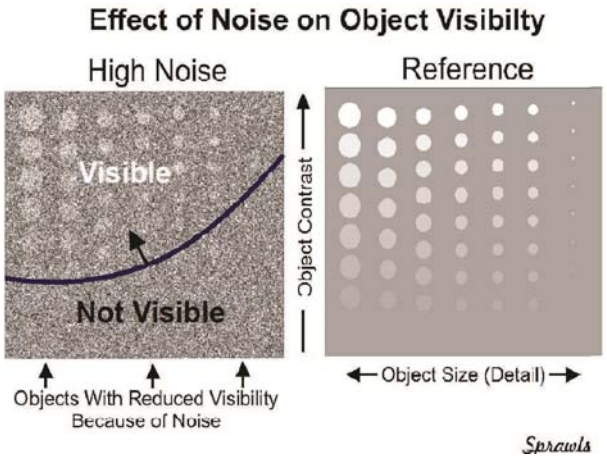


Figure 29. Using a Contrast Detail phantom to evaluate the effect of noise on the visibility of objects.

Testing an imaging procedure with a Contrast Detail phantom evaluates the effect of all three factors--contrast sensitivity, blurring, and noise--on visibility. However, it cannot determine the individual contribution of each of the factors to reduced visibility. Also, it does not provide one numerical value or score to represent the quality of an image.

Physicists measure the effect of blurring in an imaging procedure with the test device shown in Figure 30.

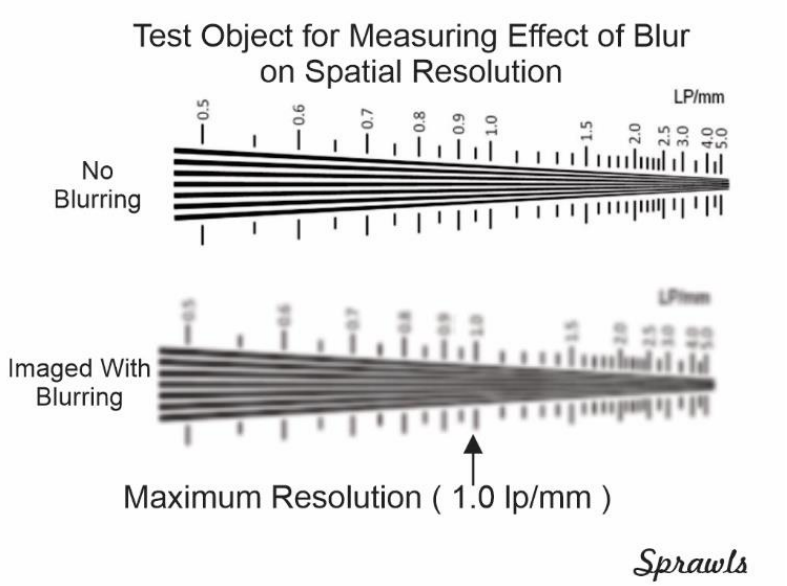


Figure 30. A test device with varying spatial resolution (size and distance between opaque lines) expressed in line pairs (a line and a space) per millimeter (mm) of distance.

Blurring reduces the visibility of the separation or resolution of the lines as shown. The highest spatial frequency that can be viewed as separated or resolved lines is the maximum resolution that provides a numerical value for the test.

Mammography is a procedure that is tested frequently by physicists to ensure that the image quality is adequate for detecting and diagnosing cancers. The specific test phantom for this is shown in Figure 31.

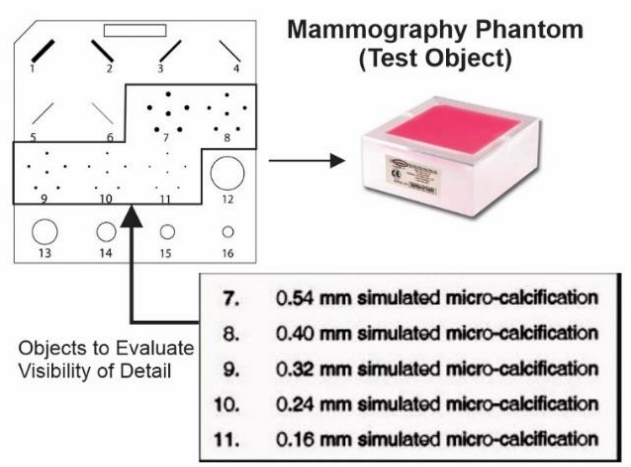
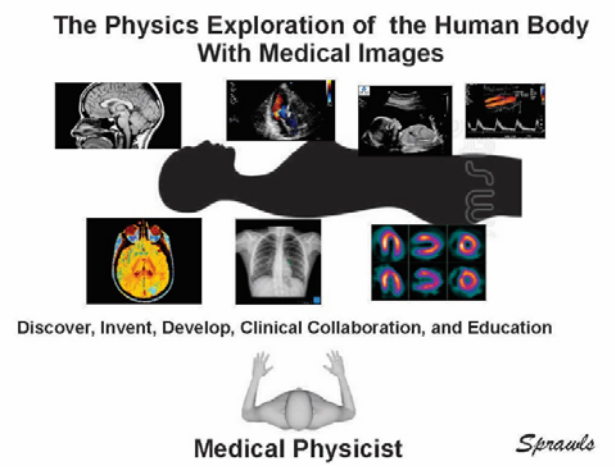


Figure 31. A test phantom specific for mammography that is used by physicists to periodically evaluate the performance of mammography system.

The phantom contains a series of objects varying in size and physical contrast, simulating objects and structures within a breast. A test is performed by producing an image and determining which objects are visible. This is a test that is required in many countries.

There are phantoms designed for the specific modalities, MRI, CT, etc., that are used by physicists in one of their major activities, ensuring the diagnostic quality of medical imaging procedures.

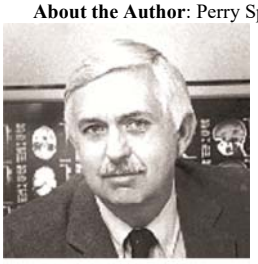
XII. THE MEDICAL PHYSICIST AND MEDICAL IMAGING



The human body is a complex physical universe that is, except for the skin surface, out of view with normal human vision. The ability to examine the interior of the body is critical to providing medical care, especially the detection and diagnosis of many major diseases and injuries. Before the 1880s the only significant access to the interior of the body for medical purposes was with surgery…cutting into the body. Then, in 1885 a major evolution in medicine occurred when the physicist, Wilhelm Roentgen discovered, researched, and demonstrated the use of a "new kind of ray", x-radiation, to produce images of the interior of the human body. This can be considered as the origin of the *medical physics profession* as we know it today. Medical imaging is a branch of physics involving the physical interactions of radiation and energy with the physical structures of the human body to produce images. Over the years many physicists along with other professionals have continued the development and evolution of medical imaging to expand its value within the practice of modern medicine.

In addition to research and development activities medical physicists are highly respected professionals working in collaboration with physicians, especially radiologists, in providing effective and safe imaging procedures and as educators.

About the Author: Perry Sprawls is a clinical medical physicist specializing in diagnostic radiology and medical physics education. He is Distinguished Emeritus Professor at Emory University School of Medicine in Atlanta and now contributes to medical physics education around the world through the Sprawls Educational Foundation, www.sprawls.org. It is the combination of his experience as a clinical physicist and educator that is the foundation for developing and sharing resources to support the teaching of medical physics. His continuing research and development activities are resulting in models for increasing the effectiveness of both the learning and teaching process, especially for clinically applied medical physics.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Magnetic Resonance Imaging

Principles, Methods, and Techniques

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The power of knowledge is realized when it is shared with others and used to benefit all people with improved health and quality of life.

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Preface

Magnetic resonance imaging (MRI) is a major medical diagnostic tool. It makes it possible to visualize and analyze a variety of tissue characteristics, blood flow and distribution, and several physiologic and metabolic functions. Much of this power comes from the ability to adjust the imaging process to be especially sensitive to each of the characteristics being evaluated. Think of it as a multipurpose imaging and analytical procedure that can be configured and optimized to provide answers to a wide range of clinical questions for virtually all parts and systems of the body.

Each imaging procedure is guided by a protocol consisting of a selection from a choice of imaging methods, selected values for a large number of imaging parameters or factors associated with the specific method, and the application of a variety of techniques to optimize image quality and acquire the images in the shortest time consistent with other procedure requirements.

Even though many imaging protocols are preprogrammed into modern MRI systems, maximum performance and benefit requires a highly educated and trained staff to conduct the procedures and to interpret the results.

The physicians who are requesting, supervising, and interpreting the MR examinations require knowledge of MRI principles, methods, and techniques to select appropriate imaging methods and techniques and to understand the basis for the clinical information conveyed in the images.

The technologists who perform the examinations need a good knowledge and understanding of the total process so that they can select and modify protocols as necessary, monitor and optimize image quality, and provide for patient and staff safety.

Medical physicists who provide support to the clinical activities with respect to image quality and procedure optimization, and conduct educational activities for other medical professionals must also have a broad knowledge at the practical and applied levels.

This book is designed to meet the needs of all who play a role in the MR imaging process. First, it develops the very important concepts of the physical principles on which MR imaging is based. It then builds an understanding of the various methods and techniques that are the heart of each imaging procedure. It gives special emphasis to image quality and the associated issues of optimizing protocols. Safety concerns are addressed in order to have an informed staff who can take a realistic approach to reducing risk and increasing patient comfort and acceptance.

The objective of this book is to help all of us obtain maximum performance and benefit from the advanced and sophisticated MR technology that is available today. Humans with the knowledge of how to apply the various imaging options to the wide range of clinical needs is, and will continue to be, a vital link in the total MR imaging process.

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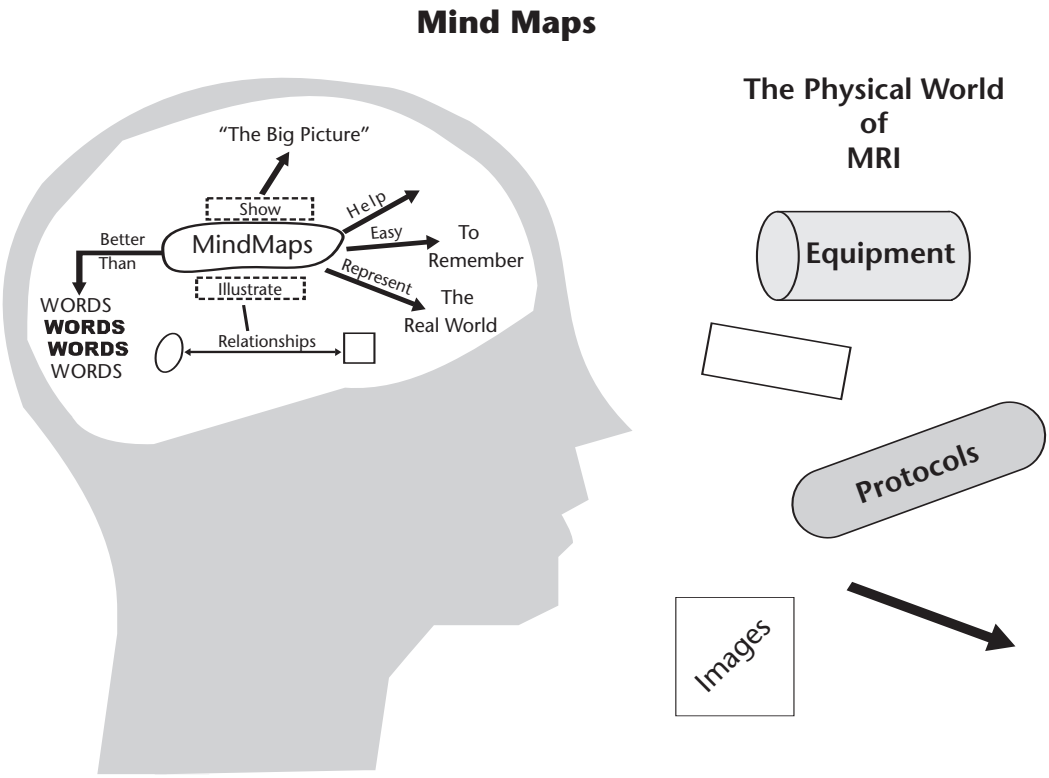
Mind Maps



MRI is not a difficult subject to learn. It is, however, a somewhat complex topic because there are many different parts of the total MR process. A useful knowledge of MRI does not consist of memorized definitions and facts. It consists of a visual representation of the MR process in our minds. It is this type of knowledge that we can use to produce and understand images. This book is designed to help you develop the concepts of the various aspects of MRI and to see how these concepts can be applied to practical imaging situations.

Each chapter is summarized with a mind map to help organize the many individual concepts. A mind map is an effective method for showing relationships and developing a comprehensive mental picture of a topic. It is a good way to provide some organization to the complexity of a topic such as MRI. Mind maps are excellent tools for study and review. The best mind maps are often the ones that each of us develops for our own use. You are encouraged to add to and modify the printed maps and to develop mind maps of your own.

1

Magnetic Resonance Image Characteristics

Introduction And Overview

Magnetic resonance imaging (MRI) is a medical imaging process that uses a magnetic field and radio frequency (RF) signals to produce images of anatomical structures, of the presence of disease, and of various biological functions within the human body. MRI produces images that are distinctly different from the images produced by other imaging modalities. A primary difference is that the MRI process can selectively image several different tissue characteristics. A potential advantage of this is that if a pathologic process does not alter one tissue characteristic and produce contrast, it might be visible in an image because of its effect on other characteristics. This causes the MRI process to be somewhat more complex than most imaging methods. In order to optimize an MRI procedure for a specific clinical examination, the user must have a good knowledge of the characteristics of the magnetic resonance (MR) image and how those characteristics can be controlled.

In this chapter we will develop a basic knowledge and overview of the MR image, how the image relates to specific tissue characteristics, and how image quality characteristics can be controlled.

The MR Image

The MR image displays certain physical characteristics of tissue. Let us now use Figure 1-1 to identify these characteristics and to see how they are related.

The MR image is a display of RF signals that are emitted by the tissue during the image acquisition process. The source of the signals is a condition of magnetization that is produced in the tissue when the patient is placed in the strong magnetic field. The tissue magnetization depends on the presence of magnetic nuclei. The specific physical characteristic of tissue or fluid that is visible in the image depends on how the magnetic field is being changed during the acquisition process. An image acquisition consists of an acquisition cycle, like a heartbeat, that is repeated many times. During each cycle the tissue magnetization is forced through a series of changes. As we will soon learn in much more detail, all tissues and fluids do not progress through these changes at the same rate. It is the level of magnetization that is present at a special "picture snapping time" at the end of each cycle that determines the intensity of the RF signal produced and the resulting tissue brightness in the image.

MR images are generally identified with specific tissue characteristics or blood conditions

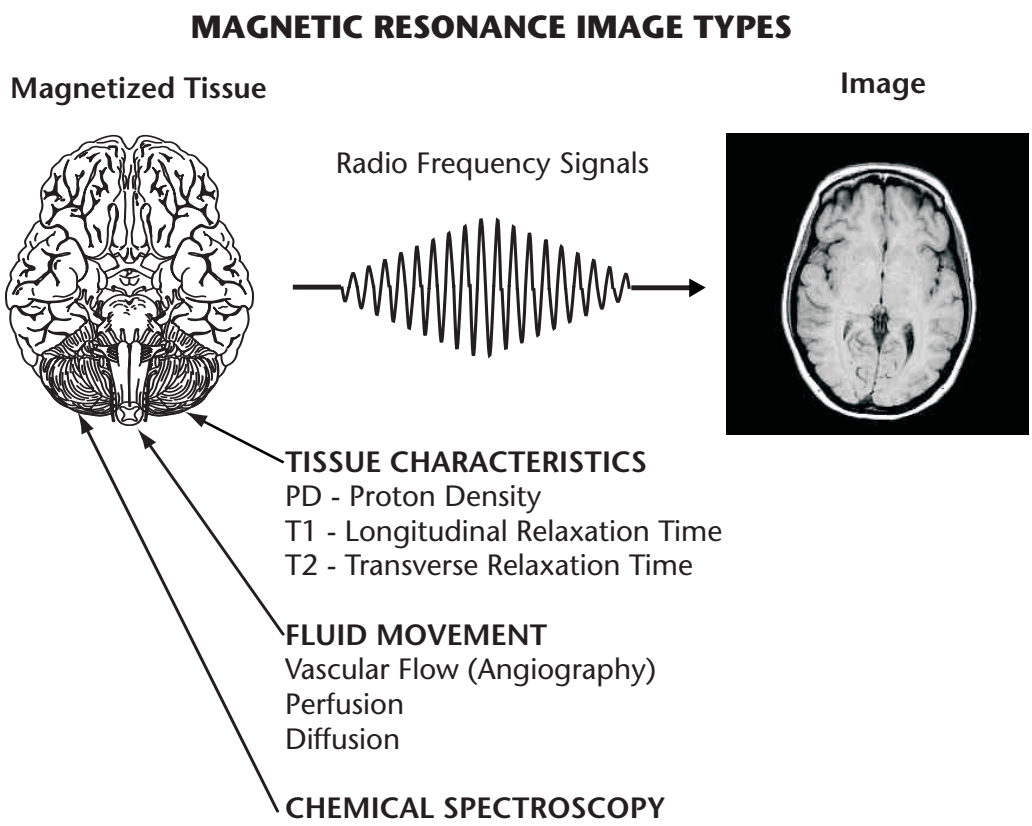


Figure 1-1. Physical characteristics of tissue and fluid movement that can be displayed in the magnetic resonance image. MRI can also provide certain chemical information by applying spectroscopy analysis to the RF signals emitted by the tissue.

that are the predominant source of contrast. These characteristics determine the level of tissue magnetization and contrast present at the time the "picture is snapped." The equipment operator, who sets the imaging protocol, determines the type of image that is to be produced by adjusting various imaging factors.

The characteristics that can be used as a source of image contrast fall into three rather distinct categories. The first, and most widely used, category is the magnetic characteristics of tissues. The second category is characteristics of fluid (usually blood) movement. The third category is the spectroscopic effects related to molecular structure.

At this time we will briefly introduce each of these characteristics to set the stage for the much more detailed descriptions presented later.

Tissue Characteristics and Image Types

Proton Density (PD) Images

The most direct tissue characteristic that can be imaged is the concentration or density of protons (hydrogen). In a proton density image the tissue magnetization, RF signal intensity, and image brightness are determined by the proton (hydrogen) content of the tissue. Tissues that are rich in protons will produce strong signals and have a bright appearance.

Magnetic Relaxation Times — T1 and T2 Images

During an MRI procedure the tissue magnetization is cycled by flipping it into an unstable condition and then allowing it to recover. This recovery process is known as *relaxation*. The time required for the magnetization to relax varies from one type of tissue to another. The relaxation times can be used to distinguish (i.e., produce contrast) among normal and pathologic tissues.

Each tissue is characterized by two relaxation times: Tl and T2. Images can be created in which either one of these two characteristics is the predominant source of contrast. It is usually not possible to create images in which one of the tissue characteristics (e.g., PD, T1, or T2) is the only pure source of contrast. Typically, there is a mixing or blending of the characteristics but an image will be more heavily *weighted* by one of them. When an image is described as a T1-weighted image, this means that T1 is the predominant source of contrast but there is also some possible contamination from the PD and T2 characteristics.

Fluid Movement and Image Types

Vascular Flow

The MRI process is capable of producing images of flowing blood without the use of contrast media. Although flow effects are often visible in all types of images, it becomes the predominant source of contrast in images produced specifically for vascular or angiographic examinations as described in Chapter 12.

Perfusion and Diffusion

It is possible to produce images that show both perfusion and diffusion within tissue. These require specific imaging methods and are often characterized as functional imaging.

Spectroscopic and Chemical Shift

The frequency of the RF signals emitted by tissue is affected to a small degree by the size and characteristics of the molecules containing the magnetic nuclei. These differences in frequencies, the chemical shift, can be displayed in images. It is also the basis of MR spectroscopy. Spectroscopy is the process of

using magnetic resonance to analyze the chemical composition of tissue. Spectroscopy makes use of the fact that different molecular structures have different resonant frequencies. Typically, the MR signals from a tissue specimen are sorted and displayed on a frequency scale. The signals from different chemical compounds will appear as peaks along the frequency scale. This leads to their identity and measure of relative abundance.

What Do You See In An MR Image?

We have discovered that an MR image can display a variety of tissue and body fluid characteristics. However, there are several physical characteristics that form the link between the image and the tissue characteristics described above. Understanding this link gives us a better appreciation of how the tissue characteristics are made visible. We will use Figure 1-2 to develop the link.

Radio Frequency Signal Intensity

The first thing we see in an image is RF signal intensity emitted by the tissues. Bright areas in the image correspond to tissues that emit high signal intensity. There are also areas in an image that appear as dark voids because no signals are produced. Between these two extremes there will be a range of signal intensities and shades of gray that show contrast or differences among the various tissues.

Let us now move deeper into the imaging process and discover the relationship between RF signal intensity and other characteristics.

Tissue Magnetization

The condition within the tissue that produces the RF signal is *magnetization*. At this point we will use an analogy to radioactive nuclide imaging. In nuclear medicine procedures it is the presence of radioactivity in the tissues that produces the radiation. In MRI it is the magnetization within the tissues that produces the RF signal radiation displayed in the image. Therefore, when we look at an MR image, we are seeing a display of magnetized tissue.

We will soon discover that tissue becomes magnetized when the patient is placed in a strong magnetic field. However, all tissues are not magnetized to the same level. During the imaging process the tissue magnetization is cycled through a series of changes, but all tissues do not change at the same rate. It is this difference in rates of change of the magnetization that makes the tissues different and produces much of the useful contrast. This will be described in much more detail later when we will learn that these rates of change are described as magnetic relaxation times, T1 and T2.

It is the level of magnetization at specific "picture snapping" times during the imaging procedure that determines the intensity of the resulting RF signal and image brightness. The MR image is indeed an image of magnetized tissue. Tissues or other materials that are not adequately magnetized during the imaging procedure will not be visible in the image.

Protons (Magnetic Nuclei)

The next thing we see is an image of protons that are the nuclei of hydrogen atoms. That is why an MRI procedure is often referred to as proton imaging.

The magnetization of tissue, which produces the RF signals, comes from protons that are actually small magnets (magnetic nuclei) present in the tissue. These small magnets are actually the nuclei of certain atoms that have a special magnetic property called a *magnetic moment*. Not all chemical substances have an adequate abundance of magnetic nuclei.

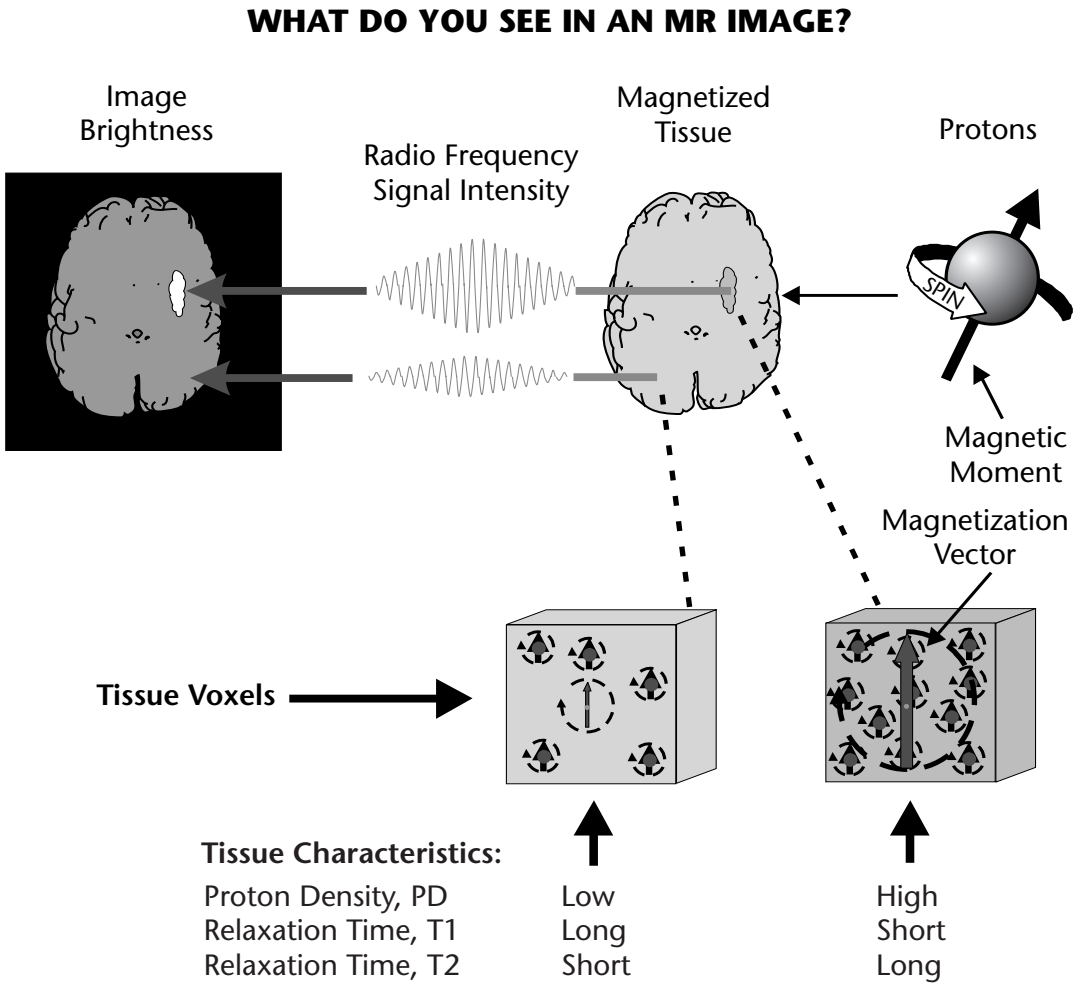


Figure 1-2. The physical characteristics that form the link between the image and the three tissue characteristics.

Hydrogen

The only substance found in tissue that has an adequate concentration of magnetic nuclei to produce good images is hydrogen. The nucleus of a hydrogen atom is a single proton. Therefore, the MR image is an image of hydrogen. When tissue that contains hydrogen (small magnetic nuclei), i.e., protons, is placed in a strong magnetic field, some of the protons line up in the same direction as the magnetic field. This alignment produces the magnetization in the tissue, which then produces the RF signal. If a tissue does not have an adequate concentration of molecules containing hydrogen, it will not be visible in an MR image.

Tissue Characteristics

As we have moved deeper into the imaging process we arrive again at the three tissue characteristics: PD, T1, and T2. It is these characteristics that we want to see because they give us valuable information about the

tissues. These characteristics become visible because each one has an effect on the level of magnetization that is present at the picture snapping time in each imaging cycle. At this time we will briefly describe the effect of each and then develop the process in more detail in Chapters 4 and 5.

PD (Proton Density)

PD has a very direct effect on tissue magnetization and the resulting RF signal and image brightness. That is because the magnetization is produced by the protons. Therefore, a tissue with a high PD can reach a high level of magnetization and produce an intense signal.

T1

When the imaging protocol is set to produce a T1-weighted image, it is the tissues with the short T1 values that produce the highest magnetization and are the brightness in the image.

T2

When the imaging protocol is set to produce a T2-weighted image, it is the tissues with the long T2 values that are the brightest. This is because they have a higher level of magnetization at the picture snapping time.

Spatial Characteristics

Figure 1-3 illustrates the basic spatial characteristics of the MR image. MRI is basically a tomographic imaging process, although there are some procedures, such as angiography, in which a complete anatomical volume will be displayed in a single image. The protocol for the acquisition process must be set up to produce the appropriate spatial characteristics for a specific clinical procedure. This includes such factors as the number of slices, slice orientation, and the structure within each individual slice.

Slices

A typical examination will consist of at least one set of contiguous slices. In most cases the entire set of slices is acquired simultaneously. However, the number of slices in a set can be limited by certain imaging factors and the amount of time allocated to the acquisition process.

The slices can be oriented in virtually any plane through the patient's body. The major restriction is that images in the different planes cannot generally be acquired simultaneously. For example, if both axial and sagittal images are required, the acquisition process must be repeated. However, there is the possibility of acquiring 3-D data from a large volume of tissue and then reconstructing slices in the different planes, as will be described in Chapter 9.

Voxels

Each slice of tissue is subdivided into rows and columns of individual volume elements, or voxels. The size of a voxel has a significant effect on image quality. It is controlled by a combination of protocol factors as described in Chapter 10 and should be adjusted to an optimum size for each type of clinical examination. Each voxel is an independent source of RF signals. That is why voxel size is a major consideration in each image acquisition.

Image Pixels

The image is also divided into rows and columns of picture elements, or pixels. In general, an image pixel represents a corresponding voxel of tissue within the slice. The brightness of an image pixel is determined by the intensity of the RF signal emitted by the tissue voxel.

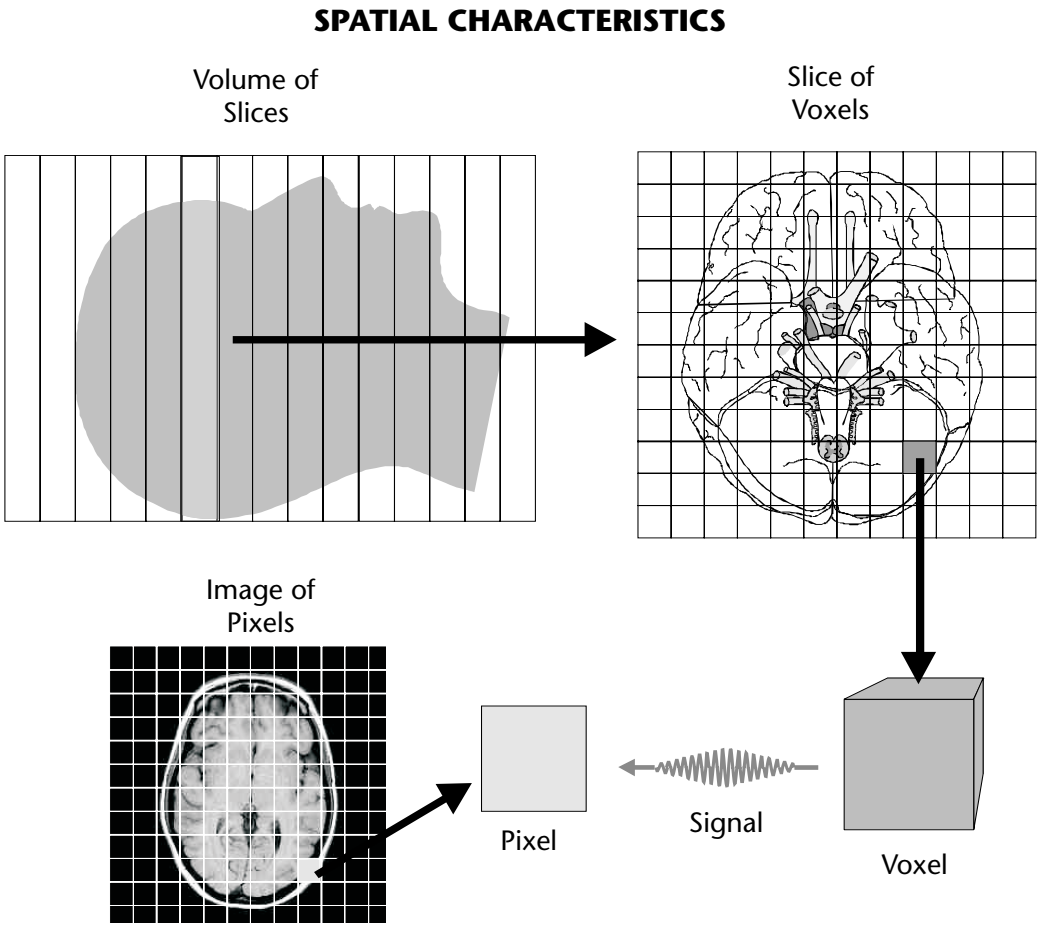


Figure 1-3. The spatial characteristics of MR images.

Control Of Image Characteristics

The operator of an MRI system has tremendous control over the characteristics and the quality of the images that are produced. The five basic image quality characteristics are represented in Figure 1-4. Each of these image characteristics is affected by a combination of the imaging factors that make up the acquisition protocol.

Not all types of clinical procedures require images with the same characteristics. Therefore, the primary objective is to use an imaging protocol in which the acquisition process is optimized for a specific clinical requirement.

Although each of the image characteristics will be considered in detail in later chapters, we will introduce them here.

Contrast Sensitivity

Contrast sensitivity is the ability of an imaging process to produce an image of objects or tissues in the body that have relatively small physical differences or inherent contrast. The contrast that is to be imaged is in the form of

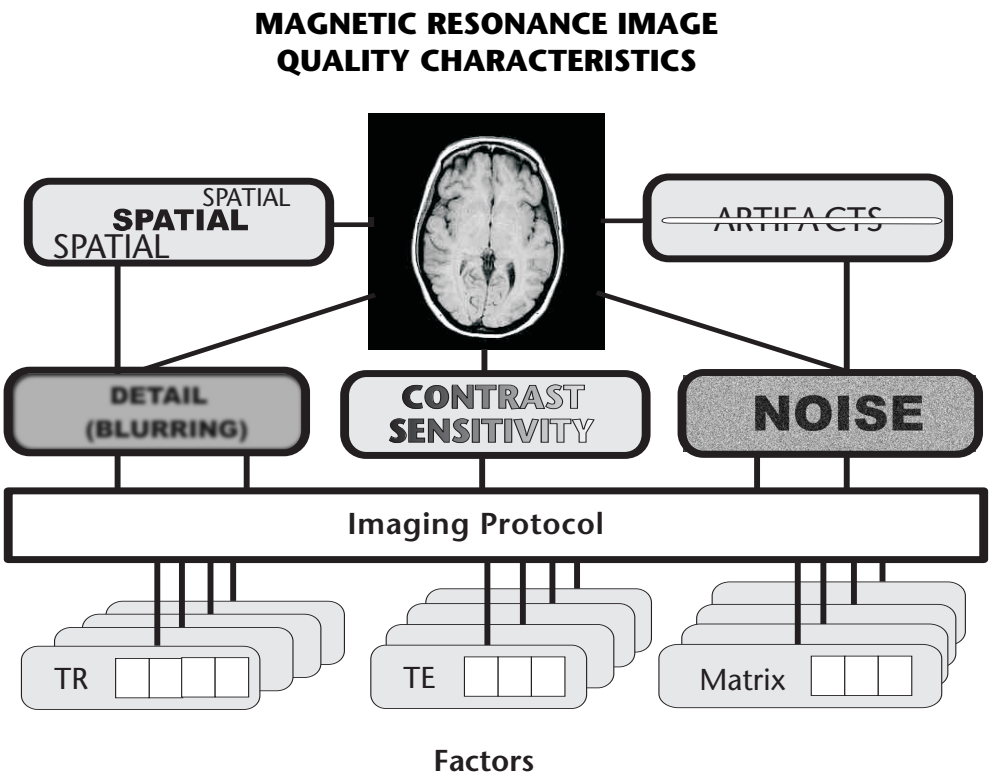


Figure 1-4. Image quality characteristics that can be controlled by the selection of protocol factors.

some specific physical characteristic. In x-ray imaging, including CT (computed tomography), difference in physical density is a principle source of contrast. One of the major advantages of MRI is that it has a high contrast sensitivity for visualizing differences among the tissues in the body because there are several sources of contrast; that is, it has the ability to image a variety of characteristics (PD, T1, T2) as described previously. Also, there is usually much greater variation among these characteristics than among the tissue density values that are the source of contrast for x-ray imaging. If a certain pathologic condition does not produce a visible change in one characteristic, there is the possibility that it will be visible by imaging some of the other characteristics.

Even though MRI has high contrast sensitivity relative to most of the other imaging modalities, it must be optimized for each clinical procedure. This includes the selection of the characteristics, or sources of contrast, that are to be imaged and then adjusting the protocol factors so that the sensitivity to that specific characteristic is optimized. This is illustrated in Figure 1-5.

Detail

A distinguishing characteristic of every imaging modality is its ability to image small objects and structures within the body. Visibility of anatomical detail (sometimes referred to as spatial resolution) is limited by the blurring that occurs during the imaging process. All medical imaging methods produce images with some

blurring but not to the same extent. The blurring in MRI is greater than in radiography. Therefore, MRI cannot image small structures that are visible in conventional radiographs.

In MRI, like all modalities, the amount of blurring and the resulting visibility of detail can be adjusted during the imaging process. Figure 1-6 shows images with different levels of blurring and visibility of detail. The protocol factors that are used to adjust detail and the associated issues in their optimization will be discussed in Chapter 10.

Noise

Visual noise is a major issue in MRI. The presence of noise in an image reduces its quality, especially by limiting the visibility of low contrast objects and differences among tissues. Figure 1-7 shows images with different levels of visual noise. Most of the noise in MR images is the result of a form of random, unwanted RF energy picked up from the patient's body.

The amount of noise can generally be controlled through a combination of factors as described in Chapter 10. However, many of these factors involve compromises with other characteristics.

Artifacts

Artifacts are undesirable objects, such as streaks and spots, that appear in images which do not directly represent an anatomical structure. They are usually produced by certain interactions of the patient's body or body functions (such as motion) with the imaging process.

There is a selection of techniques that can be used to reduce the presence of artifacts. These will be described in Chapter 14.

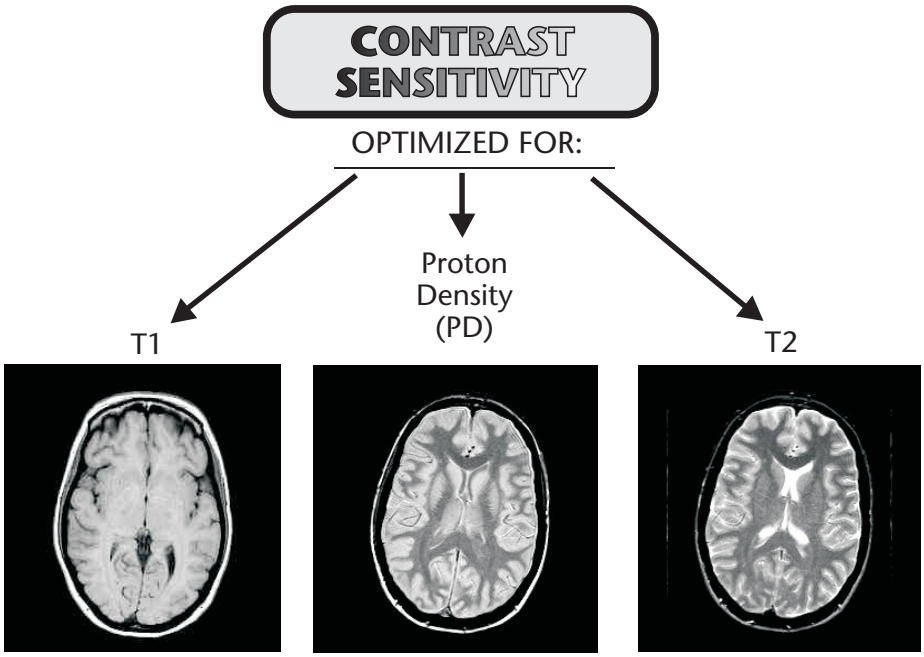


Figure 1-5. The images produced when the contrast sensitivity is optimized for each of the three specific tissue characteristics.

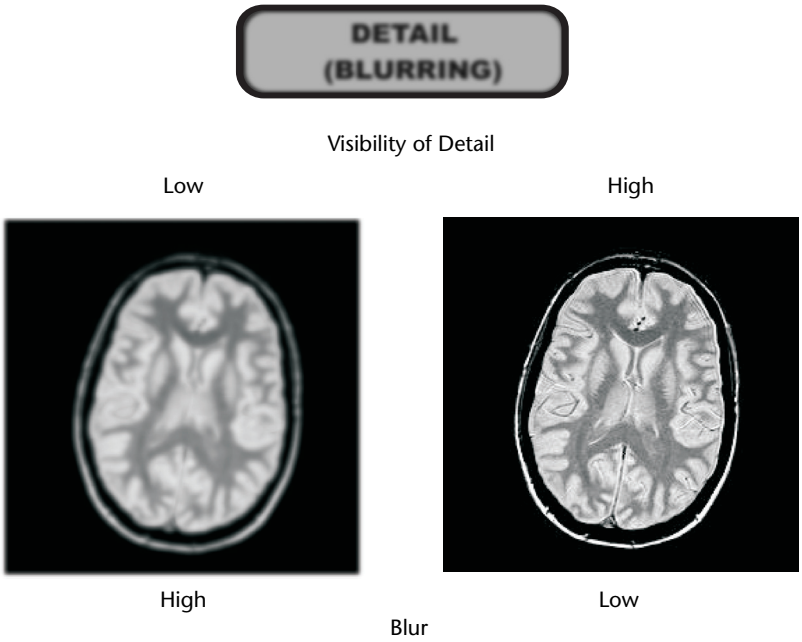


Figure 1-6. Images with different levels of blurring and visibility of anatomical detail.

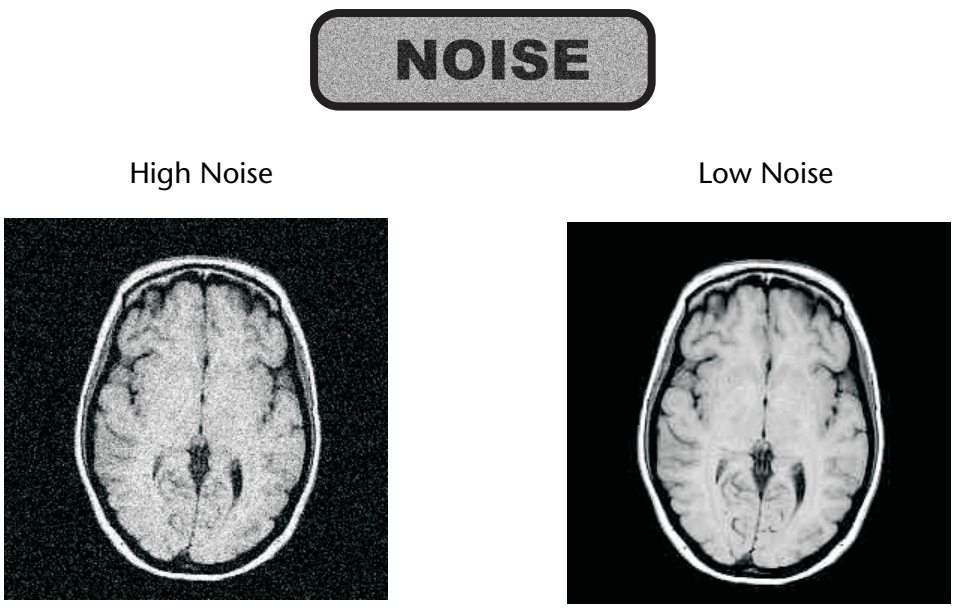


Figure 1-7. Images with different levels of visual noise.

Spatial

The general spatial characteristics of the MR image were described previously. However, when setting up an imaging protocol the spatial characteristics must be considered in the general context of image quality. As we will discover later, voxel size plays a major role in determining both image detail and image noise.

Image Acquisition Time

When considering and adjusting MR image quality, attention must also be given to the time required for the acquisition process. In general, several aspects of image quality, such as detail and noise, can be improved by using longer acquisition times.

Protocol Optimization

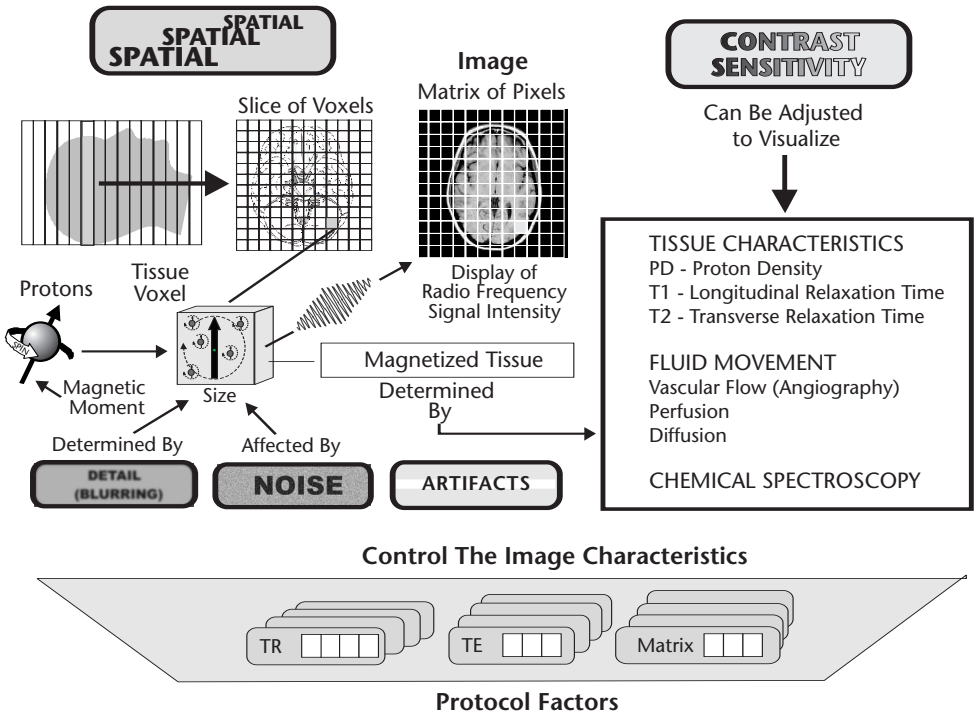
An optimum imaging protocol is one in which there is a proper balance among the image quality characteristics described above and also a balance between overall image quality and acquisition time.

The imaging protocol that is used for a specific clinical examination has a major impact on the quality of the image and the visibility of anatomical structures and pathologic conditions.

Therefore, the users of MRI must have a good knowledge of the imaging process and the protocol factors and know how to set them to optimize the image characteristics.

The overall process of optimizing protocols will be described in Chapter 11.

Mind Map Summary
Magnetic Resonance Image Characteristics



The magnetic resonance image is a display of radio frequency signal intensities that are emitted by magnetized tissue during the imaging process. The tissue becomes magnetized because it contains protons that are the magnetic nuclei of hydrogen atoms. When placed in the strong magnetic field, some of the protons align with the field producing the tissue magnetization. The level of magnetization at the time during the procedure when the "picture is snapped" is determined by a variety of tissue and fluid movement characteristics. By adjusting the imaging process it is possible to produce images in which these various characteristics are the principal sources of contrast.

An advantage of MRI is the ability to selectively image a variety of tissue and fluid characteristics. If a specific pathologic condition is not visible when viewing one characteristic, there is the possibility of seeing it by imaging some of the other characteristics.

During the imaging procedure a section of the patient's body is divided first into slices, and the slices are divided into a matrix of voxels. Each voxel is an independent RF signal source. Voxel size can be adjusted and is what determines image detail and also affects image noise.

The five major image quality characteristics—contrast sensitivity, detail, noise, artifacts, and spatial—can be controlled to a great extent by the settings of the various protocol factors.

MRI is a powerful diagnostic tool because the process can be optimized to display a wide range of clinical conditions. However, maximum benefit requires a staff with the knowledge to control the process and interpret the variety of images.

2

Magnetic Resonance Imaging System Components

Introduction And Overview

The MRI system consists of several major components, as shown in Figure 2-1. At this time we will introduce the components and indicate how they work together to create the MR image. The more specific details of the image forming process will be explained in later chapters.

The heart of the MRI system is a large magnet that produces a very strong magnetic field. The patient's body is placed in the magnetic field during the imaging procedure. The magnetic field produces two distinct effects that work together to create the image.

Tissue Magnetization

When the patient is placed in the magnetic field, the tissue becomes temporarily magnetized because of the alignment of the protons, as described previously. This is a very low-level effect that disappears when the patient is removed from the magnetic field. The ability of MRI to distinguish between different types of tissue is based on the fact that different tissues, both normal and pathologic, will become magnetized to different levels or will change their levels of magnetization (i.e., relax) at different rates.

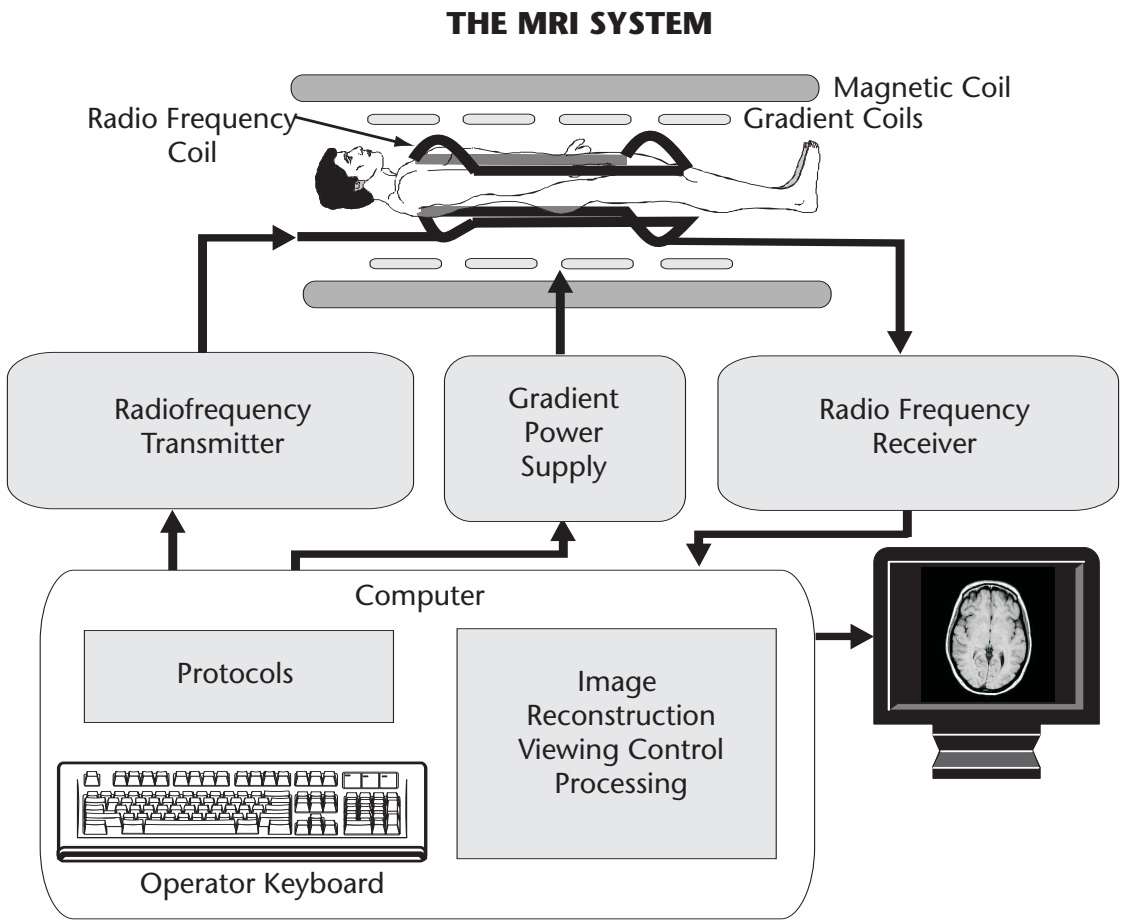


Figure 2-1. The major components of the Magnetic Resonance Imaging System.

Tissue Resonance

The magnetic field also causes the tissue to "tune in" or resonate at a very specific radio frequency. That is why the procedure is known as *magnetic resonance imaging*. It is actually certain nuclei, typically protons, within the tissue that resonate. Therefore, the more comprehensive name for the phenomenon that is the basis of both imaging and spectroscopy is *nuclear magnetic resonance* (NMR).

In the presence of the strong magnetic field the tissue resonates in the RF range. This causes the tissue to function as a tuned radio receiver and transmitter during the imaging process. The production of an MR image involves two-way radio communication between the tissue in the patient's body and the equipment.

The Magnetic Field

Figure 2-2 shows the general characteristics of a typical magnetic field. At any point within a magnetic field, the two primary characteristics are *field direction* and *field strength.*

Field Direction

It will be easier to visualize a magnetic field if it is represented by a series of parallel lines, as shown in Figure 2-2. The arrow on each line

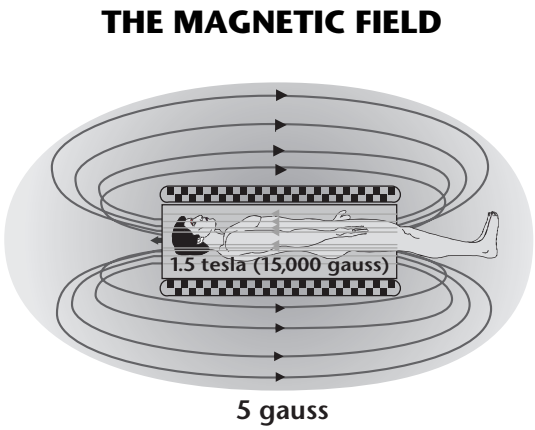


Figure 2-2. The magnetic field produced by superconducting magnets.

indicates the direction of the field. On the surface of the earth, the direction of the earth's magnetic field is specified with reference to the north and south poles. The north-south designation is generally not applied to magnetic fields used for imaging. Most of the electromagnets used for imaging produce a magnetic field that runs through the bore of the magnet and parallel to the major patient axis. As the magnetic field leaves the bore, it spreads out and encircles the magnet, creating an external fringe field. The external field can be a source of interference with other devices and is usually contained by some form of shielding.

Field Strength

Each point within a magnetic field has a particular intensity, or strength. Field strength is expressed either in the units of tesla (T) or gauss (G). The relationship between the two units is that 1.0 T is equal to 10,000 G or 10 kG. At the earth's surface, the magnetic field is relatively weak and has a strength of less than 1 G. Magnetic field strengths in the range of 0.15 T to 1.5 T are used for imaging. The significance of field strength is considered as we explore the characteristics of MR images and image quality in later chapters.

Homogeneity

MRI requires a magnetic field that is very uniform, or homogeneous with respect to strength. Field homogeneity is affected by magnet design, adjustments, and environmental conditions. Imaging generally requires a homogeneity (field uniformity) on the order of a few parts per million (ppm) within the imaging area.

High homogeneity is obtained by the process of shimming, as described later.

Magnets

There are several different types of magnets that can be used to produce the magnetic field. Each has its advantages and disadvantages.

Superconducting

Most MRI systems use superconducting magnets. The primary advantage is that a superconducting magnet is capable of producing a much stronger and stable magnetic field than the other two types (resistive and permanent) considered below. A superconducting magnetic is an electromagnet that operates in a superconducting state. A superconductor is an electrical conductor (wire) that has no resistance to the flow of an electrical current. This means that very small superconducting wires can carry very large currents without overheating, which is typical of more conventional conductors like copper. It is the combined ability to construct a magnet with many loops or turns of small wire and then use large currents that makes the strong magnetic fields possible.

There are two requirements for superconductivity. The conductor or wire must be fabricated from a special alloy and then cooled to a very low temperature. The typical magnet consists of small niobium-titanium (Nb-Ti) wires imbedded in copper. The copper has electrical resistance and actually functions as an insulator around the Nb-Ti superconductors.

During normal operation the electrical current flows through the superconductor without dissipating any energy or producing heat. If the temperature of the conductor should ever rise above the critical superconducting temperature, the current begins to produce heat and the current is rapidly reduced. This results in the collapse of the magnetic field. This is an undesirable event known as a *quench*. More details are given in Chapter 15 on safety. Superconducting magnets are cooled with liquid helium. A disadvantage of this magnet technology is that the coolant must be replenished periodically.

A characteristic of most superconducting magnets is that they are in the form of cylindrical or solenoid coils with the strong field in the internal bore. A potential problem is that the relatively small diameter and the long bore produce claustrophobia in some patients. Superconducting magnetic design is evolving to more open patient environments to reduce this concern.

Resistive

A resistive type magnet is made from a conventional electrical conductor such as copper. The name "resistive" refers to the inherent electrical resistance that is present in all materials except for superconductors. When a current is passed through a resistive conductor to produce a magnetic field, heat is also produced. This limits this type of magnet to relatively low field strengths.

Permanent

It is possible to do MRI with a non-electrical permanent magnet. An obvious advantage is that a permanent magnet does not require either electrical power or coolants for operation. However, this type of magnet is also limited to relatively low field strengths.

Both resistive and permanent magnets are usually designed to produce vertical magnetic fields that run between the two magnetic poles, as shown in Figure 2-3. Possible advantages include a more open patient environment and less external field than superconducting magnets.

Gradients

When the MRI system is in a resting state and not actually producing an image, the magnetic field is quite uniform or homogeneous over the region of the patient's body. However, during the imaging process the field must be distorted with gradients. A gradient is just a

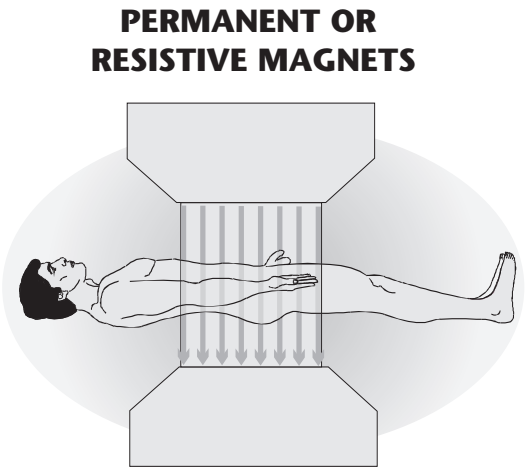


Figure 2-3. The magnetic field produced by typical resistive or permanent magnets.

change in field strength from one point to another in the patient's body. The gradients are produced by a set of gradient coils, which are contained within the magnet assembly. During an imaging procedure the gradients are turned on and off many times. This action produces the sound or noise that comes from the magnet.

The effect of a gradient is illustrated in Figure 2-4. When a magnet is in a "resting state," it produces a magnetic field that is uniform or homogenous over most of the patient's body. In this condition there are no gradients in the field. However, when a gradient coil is turned on by applying an electric current, a gradient or variation in field strength is produced in the magnetic field.

Gradient Orientation

The typical imaging magnet contains three separate sets of gradient coils. These are oriented so that gradients can be produced in the three orthogonal directions (often designated as the x, y, and z directions). Also, two or more of the gradient coils can be used together to produce a gradient in any desired direction.

Gradient Functions

The gradients are used to perform many different functions during the image acquisition process. It is the gradients that create the spatial characteristics by producing the slices and voxels that will be described in Chapter 9. The entire family of gradient echo imaging

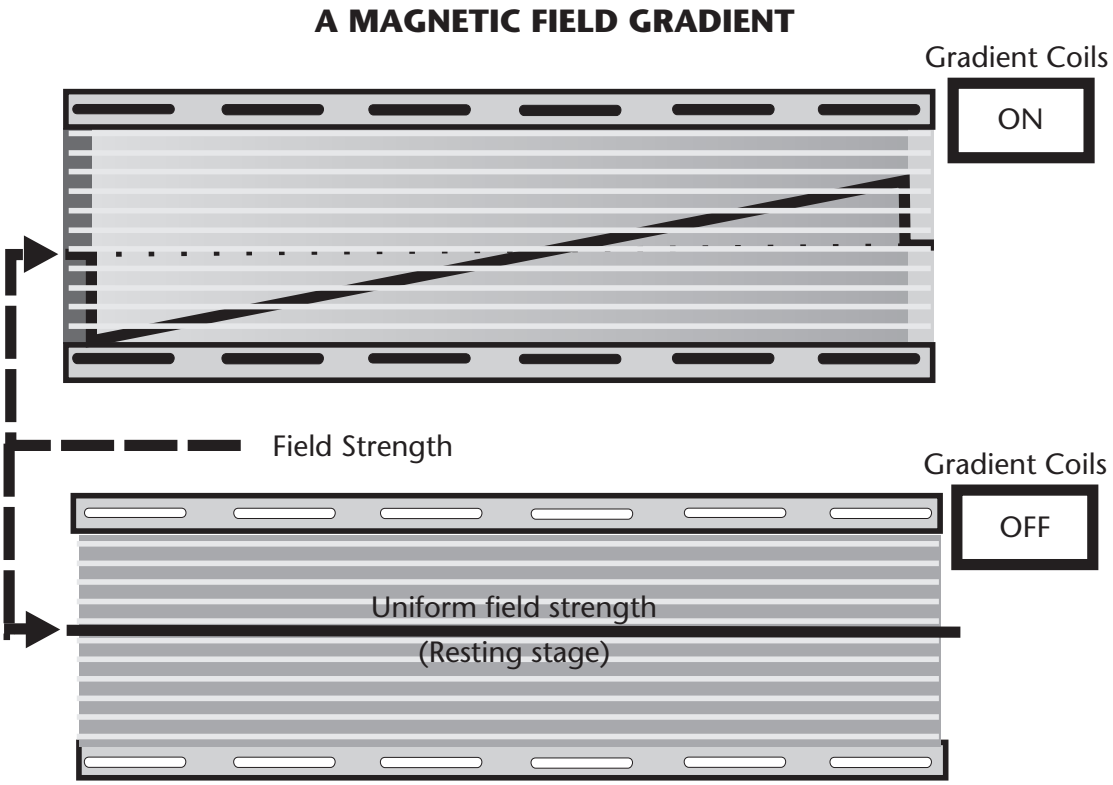


Figure 2-4. A magnetic field gradient produced by a current in the gradient coil.

methods uses a gradient to produce the echo event and signal which will be described in Chapter 7. Gradients are also used to produce one type of image contrast (phase contrast angiography) for vascular imaging, as will be described in Chapter 12, and in the functional imaging methods described in Chapter 13. Gradients also are used as part of some of the techniques to reduce image artifacts, as will be described in Chapter 14.

Gradient Strength

The strength of a gradient is expressed in terms of the change in field strength per unit of distance. The typical units are millitesla per meter (mT/m). The maximum gradient strength that can be produced is a design characteristic of a specific imaging system. High gradient strengths of 20 mT/m or more are required for the optimum performance of some imaging methods.

Risetime and Slew-Rate

For certain functions it is necessary for the gradient to be capable of changing rapidly. The *risetime* is the time required for a gradient to reach its maximum strength. The *slew-rate* is the rate at which the gradient changes with time. For example, a specific gradient system might have a risetime of 0.20 milliseconds (msec) and a slew-rate of 100 mT/m/msec.

Eddy Currents

Eddy currents are electrical currents that are induced or generated in metal structures or conducting materials that are within a changing magnetic field. Since gradients are strong, rapidly changing magnetic fields, they are capable of producing undesirable eddy currents in some of the metal components of the magnet assembly. This is undesirable because the eddy currents create their own magnetic fields that interfere with the imaging process.

Gradients are designed to minimize eddy currents either with special gradient shielding or electrical circuits that control the gradient currents in a way that compensates for the eddy-current effects.

Shimming

One of the requirements for good imaging is a homogeneous magnet field. This is a field in which there is a uniform field strength over the image area. Shimming is the process of adjusting the magnetic field to make it more uniform.

Inhomogeneities are usually produced by magnetically susceptible materials located in the magnetic field. The presence of these materials produces distortions in the magnetic field that are in the form of inhomogeneities. This can occur in both the internal and external areas of the field. Each time a different patient is placed in the magnetic field, some inhomogeneities are produced. There are many things in the external field, such as building structures and equipment, that can produce inhomogeneities. The problem is that when the external field is distorted, these distortions are also transferred to the internal field where they interfere with the imaging process. Inhomogeneities produce a variety of problems that will be discussed later.

It is not possible to eliminate all of the sources of inhomogeneities. Therefore, shimming must be used to reduce the inhomogeneities. This is done in several ways. When a magnet is manufactured and installed, some shimming might be done by placing metal shims in appropriate locations. Magnets also contain a set of shim coils. Shimming is produced by adjusting the electrical currents in these coils. General shimming is done by the

engineers when a magnet is installed or serviced. Additional shimming is done for individual patients. This is often done automatically by the system.

Magnetic Field Shielding

The external magnetic field surrounding the magnet is the possible source of two types of problems. One problem is that the field is subject to distortions by metal objects (building structures, vehicles, etc.) as described previously. These distortions produce inhomogeneities in the internal field. The second problem is that the field can interfere with many types of electronic equipment such as imaging equipment and computers.

It is a common practice to reduce the size of the external field by installing shielding as shown in Figure 2-5. The principle of magnetic field shielding is to provide a more attractive return path for the external field as it passes from one end of the magnetic field to the other. This is possible because air is not a good magnetic field conductor and can be replaced by more conductive materials, such as iron. There are two types of shielding: *passive* and *active*.

Passive Shielding

Passive shielding is produced by surrounding the magnet with a structure consisting of relatively large pieces of ferromagnetic materials such as iron. The principle is that the

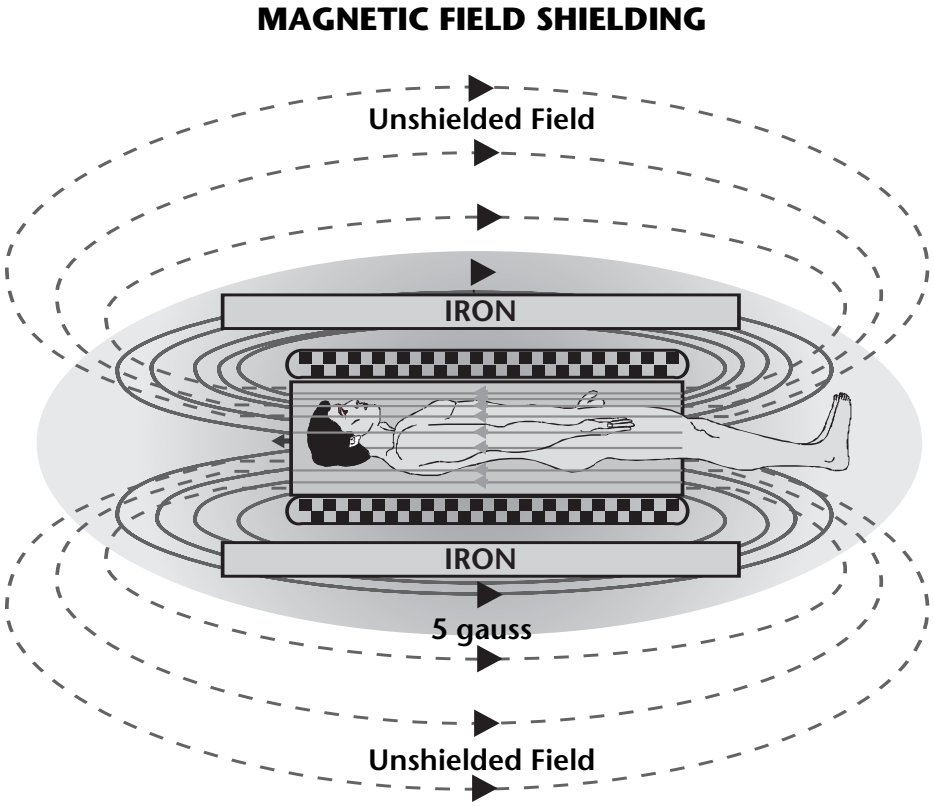


Figure 2-5. The principle of magnetic field shielding.

ferromagnetic materials are a more attractive path for the magnetic field than the air. Rather than expanding out from the magnet, the magnetic field is concentrated through the shielding material located near the magnet as shown in Figure 2-5. This reduces the size of the field.

Active Shielding

Active shielding is produced by additional coils built into the magnet assembly. They are designed and oriented so that the electrical currents in the coils produce magnetic fields that oppose and reduce the external magnetic field.

The Radio Frequency System

The radio frequency (RF) system provides the communications link with the patient's body for the purpose of producing an image. All medical imaging modalities use some form of radiation (e.g., x-ray, gamma-ray, etc.) or energy (e.g., ultrasound) to transfer the image from the patient's body.

The MRI process uses RF signals to transmit the image from the patient's body. The RF energy used is a form of non-ionizing radiation. The RF pulses that are applied to the patient's body are absorbed by the tissue and converted to heat. A small amount of the energy is emitted by the body as signals used to produce an image. Actually, the image itself is not formed within and transmitted from the body. The RF signals provide information (data) from which the image is reconstructed by the computer. However, the resulting image is a display of RF signal intensities produced by the different tissues.

RF Coils

The RF coils are located within the magnet assembly and relatively close to the patient's body. These coils function as the antennae for both transmitting signals to and receiving signals from the tissue. There are different coil designs for different anatomical regions (shown in Figure 2-6). The three basic types are body, head, and surface coils. The factors leading to the selection of a specific coil will be considered in Chapter 10. In some applications the same coil is used for both transmitting and receiving; at other times, separate transmitting and receiving coils are used.

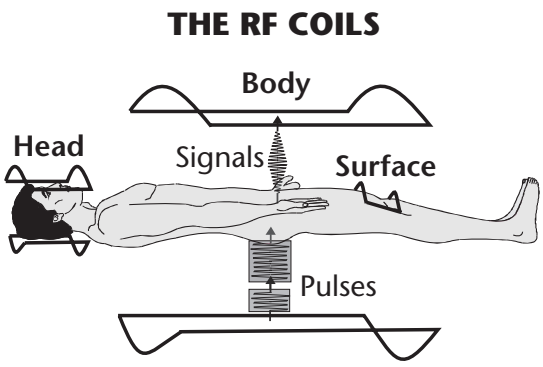


Figure 2-6. The three types of RF coils (body, head, and surface) that are the antennae for transmitting pulses and receiving signals from the patient's body.

Surface coils are used to receive signals from a relatively small anatomical region to produce better image quality than is possible with the body and head coils. Surface coils can be in the form of single coils or an array of several coils, each with its own receiver circuit operated in a *phased array* configuration. This configuration produces the high image quality obtained from small coils but with the added advantage of covering a larger anatomical region and faster imaging.

Transmitter

The RF transmitter generates the RF energy, which is applied to the coils and then transmitted to the patient's body. The energy is generated as a series of discrete RF pulses. As

we will see in Chapters 6, 7, and 8, the characteristics of an image are determined by the specific sequence of RF pulses.

The transmitter actually consists of several components, such as RF modulators and power amplifiers, but for our purposes here we will consider it as a unit that produces pulses of RF energy. The transmitters must be capable of producing relatively high power outputs on the order of several thousand watts. The actual RF power required is determined by the strength of the magnetic field. It is actually proportional to the square of the field strength. Therefore, a 1.5 T system might require about nine times more RF power applied to the patient than a 0.5 T system. One important component of the transmitter is a power monitoring circuit. That is a safety feature to prevent excessive power being applied to the patient's body, as described in Chapter 15.

Receiver

A short time after a sequence of RF pulses is transmitted to the patient's body, the resonating tissue will respond by returning an RF signal. These signals are picked up by the coils and processed by the receiver. The signals are converted into a digital form and transferred to the computer where they are temporarily stored.

RF Polarization

The RF system can operate either in a linear or a circularly polarized mode. In the circularly polarized mode, quadrature coils are used. Quadrature coils consist of two coils with a 90˚ separation. This produces both improved excitation efficiency by producing the same effect with half of the RF energy (heating) to the patient, and a better signal-to-noise ratio for the received signals.

RF Shielding

RF energy that might be in the environment could be picked up by the receiver and interfere with the production of high quality images. There are many sources of stray RF energy, such as fluorescent lights, electric motors, medical equipment, and radio communications devices. The area, or room, in which the patient's body is located must be shielded against this interference.

An area can be shielded against external RF signals by surrounding it with an electrically conducted enclosure. Sheet metal and copper screen wire are quite effective for this purpose.

The principle of RF shielding is that RF signals cannot enter an electrically conductive enclosure. The thickness of the shielding is not a factor—even thin foil is a good shield. The important thing is that the room must be completely enclosed by the shielding material without any holes. The doors into imaging rooms are part of the shielding and should be closed during image acquisition.

Computer Functions

A digital computer is an integral part of an MRI system. The production and display of an MR image is a sequence of several specific steps that are controlled and performed by the computer.

Acquisition Control

The first step is the acquisition of the RF signals from the patient's body. This acquisition process consists of many repetitions of an imaging cycle. During each cycle a sequence of RF pulses is transmitted to the body, the gradients are activated, and RF signals are collected. Unfortunately, one imaging cycle does not produce enough signal data to create an image. Therefore, the imaging cycle must be repeated many times to form an image. The time required to acquire images is determined by the duration of the imaging cycle or cycle repetition time—an adjustable factor known

as TR—and the number of cycles. The number of cycles used is related to image quality. More cycles generally produce better images. This will be described in much more detail in Chapters 10 and 11.

Protocols stored in the computer control the acquisition process. The operator can select from many preset protocols for specific clinical procedures or change protocol factors for special applications.

Image Reconstruction

The RF signal data collected during the acquisition phase is not in the form of an image. However, the computer can use the collected data to create or "reconstruct" an image. This is a mathematical process known as a Fourier transformation that is relatively fast and usually does not have a significant effect on total imaging time.

Image Storage and Retrieval

The reconstructed images are stored in the computer where they are available for additional processing and viewing. The number of images that can be stored—and available for immediate display—depends on the capacity of the storage media.

Viewing Control and Post Processing

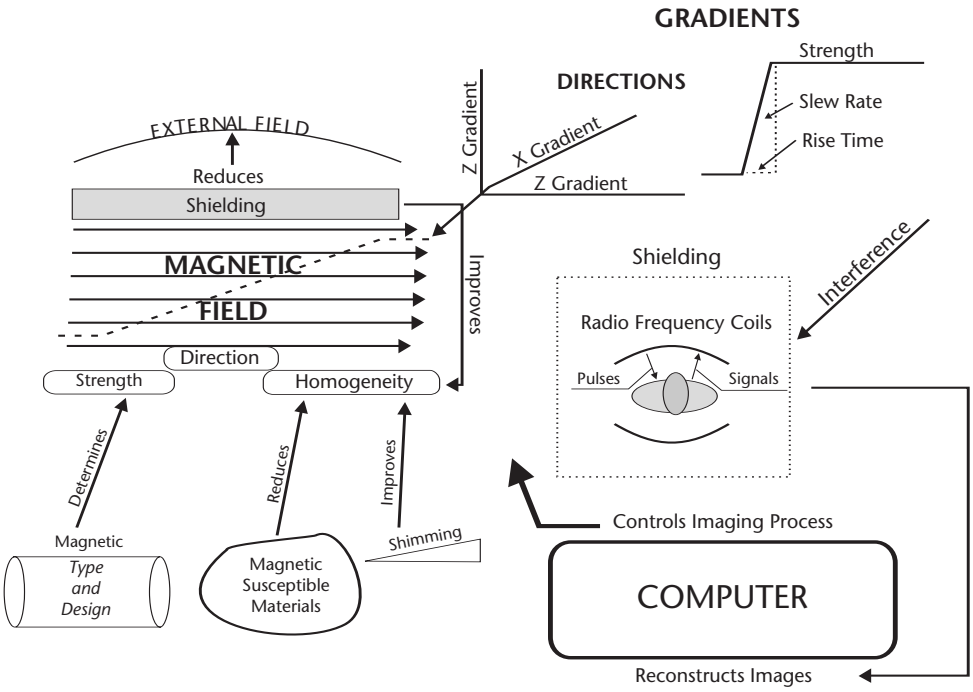
The computer is the system component that controls the display of the images. It makes it possible for the user to select specific images and control viewing factors such as windowing (contrast) and zooming (magnification).

In many applications it is desirable to process the reconstructed images to change their characteristics, to reformat an image or set of images, or to change the display of images to produce specific views of anatomical regions.

These post-processing (after reconstruction) functions are performed by a computer. In some MRI systems some of the post processing is performed on a work-station computer that is in addition to the computer contained in the MRI system.

Mind Map Summary

Magnetic Resonance Imaging System Components



The magnetic resonance imaging system consists of several major components that function together to produce images. During the image acquisition process the patient's body is placed in a strong magnetic field. At each point, the magnetic field has a specific direction. This direction is used as a reference for expressing the direction of tissue magnetization. The strength of a magnetic field is determined by the type and design of the magnet. Superconducting magnets can produce strong magnetic fields. Resistive and permanent magnets are limited to relatively weak field strengths. The homogeneity, or uniformity of field strength is necessary for good imaging. Homogeneity is reduced by magnetically susceptible materials that come into the field and produce distortions. This can occur in both the external field and within a patient's body. Shimming is the process of adjusting the magnetic field to make it more homogeneous. This can be achieved by passive shims that are added when a magnet is installed and with active shimming produced by adjusting the currents in the shimming coils.

Shielding of the magnetic field reduces the size and strength of the external magnetic field and also improves homogeneity by protecting from interference caused by objects in the external field area.

A gradient is an intentional variation in magnetic field strength that is produced by the gradient coils. There are three basic gradient coils that are oriented to produce gradients in the three

orthogonal directions. Gradients perform several functions during the image acquisition process. An important characteristic of a gradient, especially for some advanced image procedures, is its strength and how fast it can be turned on and off.

The MRI process consists of an exchange of RF pulses and signals between the equipment and the patient's body. This is done through the RF coils that serve as the antenna for transmitting the pulses and receiving the signals. It is necessary to shield the imaging area by enclosing it in a conductive metal (copper) room to block external RF interference.

The imaging process is controlled by information stored in a computer. The protocols programmed into the computer and selected by the operator guide the imaging process and determine the characteristics of the images. The RF signals collected from the patient's body during the acquisition process are used by the computer to reconstruct the image.

3

Nuclear Magnetic Resonance

Introduction And Overview

When certain materials, such as tissue, are placed in a strong magnetic field, two things happen. The materials take on a *resonant characteristic* and they become *magnetized*. In this chapter we will consider the resonant characteristic. In Chapter 4 we will study the magnetization effect. *Resonance* means the materials can absorb and then re-radiate RF radiation at a specific frequency, like a radio receiver-transmitter, as illustrated in Figure 3-1. It is actually the nuclei of the atoms that resonate. The phenomenon is generally known as nuclear magnetic resonance (NMR). The resonant frequency of material such as tissue is typically in the RF range so that the emitted radiation is in the form of radio signals. The specific resonant frequency is determined by three factors as shown in the illustration and will be described in detail later. The characteristics of the RF signals emitted by the material are determined by certain physical and chemical characteristics of the material. The RF signals produced by the NMR process can be displayed either in the form of images (MRI) or as a graph depicting chemical composition (MR spectroscopy).

Magnetic Nuclei

Materials that participate in the MR process must contain nuclei with specific magnetic properties. In order to interact with a magnetic field, the nuclei themselves must be

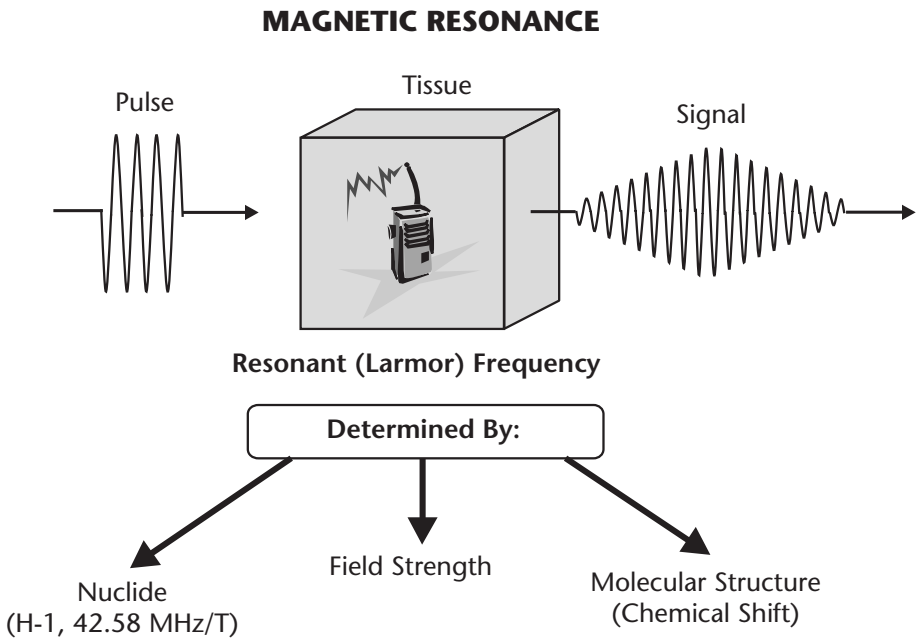


Figure 3-1. The concept of Nuclear Magnetic Resonance (NMR).

small magnets and have a magnetic property or *magnetic moment,* as shown in Figure 3-2. The magnetic characteristic of an individual nucleus is determined by its neutron-proton composition. Only certain nuclides with an odd number of neutrons and protons are magnetic. Even though most chemical elements have one or more isotopes with magnetic nuclei, the number of magnetic isotopes that might be useful for either imaging or *in vivo* spectroscopic analysis is somewhat limited. Among the nuclides that are magnetic and can participate in an NMR process, the amount of signal produced by each nuclide varies considerably.

Spins

Protons and neutrons that make up a nucleus have an intrinsic angular momentum or *spin*. Pairs of protons and neutrons align in such a way that their spins cancel. However, when there is an odd number of protons or neutrons (odd mass numbers), some of the spins will not be canceled and the total nucleus will have a net spin characteristic. It is this spinning characteristic of a particle with an electric charge (the nucleus) that produces a magnetic property known as the *magnetic moment*.

It is for this reason that magnetic nuclei, such as protons, are often referred to as *spins*.

The magnetic property, or magnetic moment, of a nucleus has a specific direction. In Figure 3-2, the direction of the magnetic moment is indicated by an arrow drawn through the nucleus.

RF Signal Intensity

The intensity of the RF signal emitted by tissue is probably the most significant factor in determining image quality and the time required to acquire an image. This important issue is considered in Chapters 10 and 11. We now begin to introduce the factors that contribute to signal intensity.

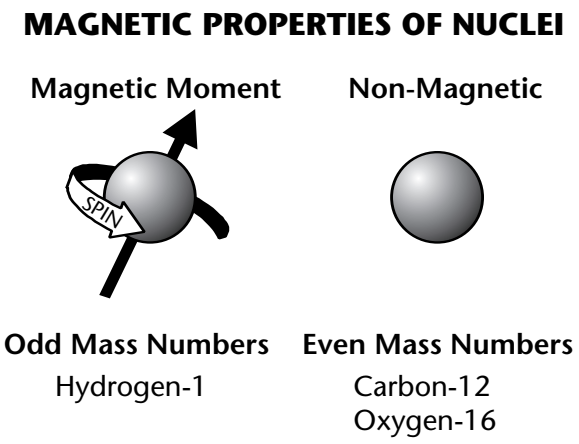


Figure 3-2. Magnetic and non-magnetic nuclei.

During the imaging process, the body section is divided into an array of individual volume elements, or voxels. It is the signal intensity from each voxel that determines image quality. The signal is produced by the magnetic nuclei within each voxel. Therefore, signal intensity is, in general, proportional to the quantity of magnetic nuclei within an individual voxel. We now consider the factors that affect the number of magnetic nuclei within an individual voxel.

Relative Signal Strength

The relative signal strength from the various chemical elements in tissue is determined by three factors: (1) tissue concentration of the element; (2) isotopic abundance; and (3) sensitivity of the specific nuclide.

In comparison to all other nuclides, hydrogen produces an extremely strong signal. This results from its high values for each of the three contributing factors.

Of the three factors, only the concentration, or density, of the nuclei varies from point to point within an imaged section of tissue. The quantity is often referred to as *proton density* and is the most fundamental tissue characteristic that determines the intensity of the RF signal from an individual voxel, and the resulting pixel brightness. In most imaging situations, pixel brightness is proportional to the density (concentration) of nuclei (protons) in the corresponding voxel, although additional factors, such as relaxation times, modify this relationship.

Protons in solids, such as the tabletop and bone, do not produce signals. Signals come only from protons in molecules that are free to move, as in a liquid state.

Tissue Concentration of Elements

The concentration of chemical elements in tissue covers a considerable range, depending on tissue type and such factors as metabolic or pathologic state. The concentrations of elements in tissue are in two groups. Four elements—hydrogen, carbon, nitrogen, and oxygen—typically make up at least 99% of tissue mass.

The most abundant isotopes of the four elements are hydrogen-1, carbon-12, nitrogen-14, and oxygen-16. Note that the mass number of hydrogen (1) is odd while the mass numbers of the other three (12, 14, 16) are even. Therefore, hydrogen is the only one of these four isotopes that has a strong magnetic nucleus. The nucleus of the hydrogen-1 atom is a single proton. Among all the chemical elements, hydrogen is unique in that it occurs in relatively high concentrations in most tissues, and the most abundant isotope (H-1) has a magnetic nucleus.

Other elements, such as sodium, phosphorus, potassium, and magnesium, are present in very low concentrations. Calcium is concentrated in bone or localized deposits.

Within this group of elements with low tissue concentrations are several with magnetic nuclei. These include fluorine-19, sodium-23, phosphorus-31, and potassium-39.

Isotopic Abundance

Most chemical elements have several isotopes. When a chemical element is found in a naturally occurring substance, such as tissue, most of the element is typically in the form of one isotope, with very low concentrations of the other isotopic forms. For the three elements—carbon, nitrogen, and oxygen—that have a high concentration in tissue, the magnetic isotopes are the ones with a low abundance in the natural state. These include carbon-l3, nitrogen-15, and oxygen-17.

Relative Sensitivity and Signal Strength

The signal strength produced by an equal quantity of the various nuclei also varies over a considerable range. This inherent NMR sensitivity is typically expressed relative to hydrogen-1, which produces the strongest signal of all of the nuclides. The relative sensitivities of some magnetic nuclides are shown in Table 3-1.

Table 3-1. Relative Sensitivities of Some Magnetic Nuclides

Nuclide	Sensitivity
Hydrogen-1	1.0
Fluorine-19	0.83
Sodium-23	0.093
Phosphorus-1	0.066

In summary, hydrogen has a lot going for it: 1) a high tissue concentration; 2) the most abundant isotope (H-1) is magnetic; and 3) it produces a relatively strong signal compared to an equal concentration of other nuclei. That is why hydrogen is the only element that is imaged with conventional MRI systems.

Radio Frequency Energy

During an imaging procedure, RF energy is exchanged between the imaging system and the patient's body. This exchange takes place through a set of coils located relatively close to the patient's body as we saw in Chapter 2. The RF coils are the antennae that transmit energy to and receive signals from the tissue.

Pulses

RF energy is applied to the body in several short pulses during each imaging cycle. The strength of the pulses is described in terms of the angle through which they rotate or flip the magnetic nuclei and the resulting tissue magnetization, as described later. Many imaging methods use both 90˚ and 180˚ pulses in each cycle.

Signals

At a specific time in each imaging cycle, the tissue is stimulated to emit an RF signal, which is picked up by the coils, analyzed, and used to form the image. The spin echo or gradient echo methods are generally used to stimulate signal emission. Therefore, the signals from the patient's body are commonly referred to as *echoes*.

Nuclear Magnetic Interactions

The NMR process is a series of interactions involving the magnetic nuclei, a magnetic field, and RF energy pulses and signals.

Nuclear Alignment

Recall that a magnetic nucleus is characterized by a magnetic moment. The direction of the magnetic moment is represented by a small arrow passing through the nucleus. If we think of the

nucleus as a small conventional magnet, the magnetic moment arrow corresponds to the south pole-north pole direction of the magnet.

In the absence of a strong magnetic field, magnetic moments of nuclei are randomly oriented in space. Many nuclei in tissue are not in a rigid structure and are free to change direction. In fact, nuclei are constantly tumbling, or changing direction, because of thermal activity within the material; in this case, tissue.

When a material containing magnetic nuclei is placed in a magnetic field, the nuclei experience a torque that encourages them to align with the direction of the field. In the human body, however, thermal energy agitates the nuclei and keeps most of them from aligning parallel to the magnetic field. The number of nuclei that do align with the magnetic field is proportional to the field strength. The magnetic fields used for imaging can align only a few of every million magnetic nuclei present. However, this is sufficient to produce a useful NMR effect.

Precession and Resonance

When a spinning magnetic nucleus aligns with a magnetic field, it is not fixed; the nuclear magnetic moment precesses, or oscillates, about the axis of the magnetic field, as shown in Figure 3-3. The precessing motion is a physical phenomenon that results from an interaction between the magnetic field and the spinning momentum of the nucleus.

Precession is often observed with a child's spinning top. A spinning top does not stand vertical for long, but begins to wobble, or precess. In this case, the precession is caused by an interaction between the earth's gravitational field and the spinning momentum of the top.

The precession rate (cycles per second) is directly proportional to the strength of the magnetic field. It is this precessing motion

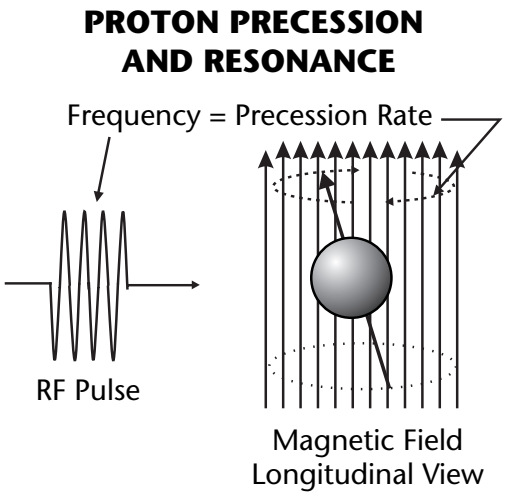


Figure 3-3. Magnetic nuclei precession and resonance in a magnetic field.

that makes a nucleus sensitive and receptive to incoming RF energy when the RF frequency matches the precession rate. This precession rate corresponds to the resonant frequency. It is the precessing nuclei, typically protons, that are tuned to receive and transmit RF energy.

Excitation

If a pulse of RF energy with a frequency corresponding to the nuclear precession rate is applied to the material, some of the energy will be absorbed by the individual nuclei. The absorption of energy by a nucleus flips its alignment away from the direction of the magnetic field, as shown in Figure 3-4. This increased energy places the nucleus in an unnatural, or *excited*, state.

In MRI an RF pulse is used that flips some of the nuclei into the transverse plane of the magnetic field. In this excited state the precession is now transformed into a spinning motion of the nucleus around the axis of the magnetic field. It should be noted that this spinning motion is an enhanced precession

and is different from the intrinsic spin of a nucleus about its own axis.

The significance of a magnetic nucleus spinning around the axis of the magnetic field is that this motion now generates an RF signal as shown in Figure 3-5. It is this signal, from many nuclei, that is collected to form the MR image.

Relaxation

When a nucleus is in an excited state, it experiences an increased torque from the magnetic field, urging it to realign. The nucleus can return to a position of alignment by transferring its excess energy to other nuclei or the general structure of the material. This process is known as *relaxation*.

Relaxation is not instantaneous following an excitation. It cannot occur until the nucleus is able to transfer its excess energy. How quickly the energy transfer takes place depends on the physical characteristics of the tissue. In fact, the nuclear relaxation rate (or time) is, in many cases, the most significant factor in producing contrast among different types of tissue in an image.

We are more interested in the collective relaxation of many nuclei that produce the magnetization of tissue and will return to this point in the next chapter.

Resonance

The significance of the nuclear precession is that it causes the nucleus to be extremely sensitive, or tuned, to RF energy that has a frequency identical with the precession frequency (rate). This condition is known as *resonance* and is the basis for all MR procedures. NMR is the process in which a nucleus resonates, or "tunes in," when it is in a magnetic field.

Resonance is fundamental to the absorption and emission of energy by many objects and devices. Objects are most effective in exchanging energy at their own resonant frequency. The resonance of an object or device

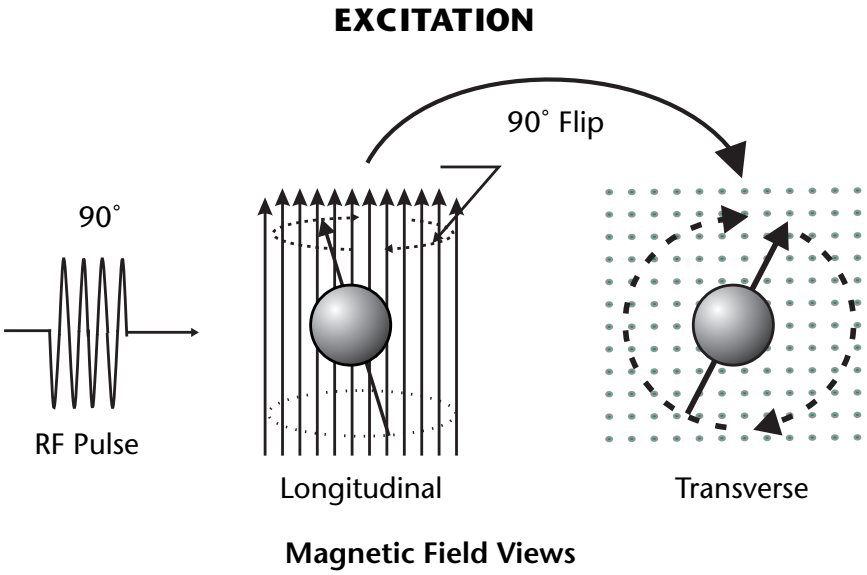


Figure 3-4. The excitation of a magnetic nucleus by the application of a pulse of RF energy.

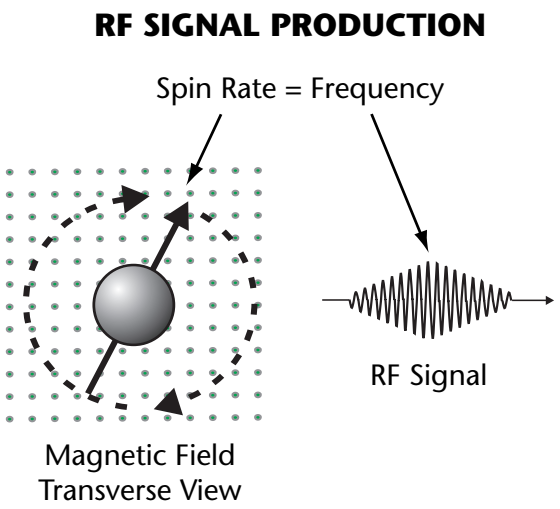


Figure 3-5. RF signal production by magnetic nuclei spinning in the transverse plane of a magnetic field.

is determined by certain physical characteristics. Let us consider two common examples.

Radio receivers operate on the principle of resonant frequency. A receiver can select a specific broadcast station because each station transmits a different frequency. Tuning a radio is actually adjusting its resonant frequency. Its receiver is very sensitive to radio signals at its resonant frequency and insensitive to all other frequencies.

The strings of a musical instrument also have specific resonant frequencies. This is the frequency at which the string vibrates to produce a specific audio frequency, or musical note. The resonant frequency of a string depends on the amount of tension. It can be changed, or tuned, by changing the tension. This is somewhat analogous to the resonant frequency of a magnetic nucleus being dependent on the strength of the magnetic field in which it is located.

Larmor Frequency

The resonant frequency of a nucleus is determined by a combination of nuclear characteristics and the strength of the magnetic field. The resonant frequency is also known as the *Larmor frequency*. The specific relationship between resonant frequency and field strength is an inherent characteristic of each nuclide and is generally designated the *gyromagnetic ratio*. The Larmor frequencies [in megahertz (MHz)] for selected nuclides in a magnetic field of 1 T are shown in Table 3-2.

Table 3-2. Larmor Frequencies for Selected Nuclides in a Magnetic Field of 1 T

Nuclide	Larmor Frequency (MHz)
Hydrogen-1	42.58
Fluorine-19	40.05
Phosphorus-31	17.24
Sodium-23	11.26

The fact that different nuclides have different resonant frequencies means that most MR procedures can "look at" only one chemical element (nuclide) at a time.

Field Strength

For all nuclides, the resonant frequency is proportional to the strength of the magnetic field. In a very general sense, increasing the magnetic field strength increases the tension on the nuclei (as with the strings of a musical instrument) and increases the resonant frequency. The fact that a specific nuclide can be tuned to different radio frequencies by varying the field strength (i.e., applying gradients) is used in the imaging process.

Chemical Shift

The resonant frequency of magnetic nuclei, such as protons, is also affected by the structure of the molecule in which they are located.

When a proton, or other magnetic nucleus, is part of a molecule, it is slightly shielded from the large magnetic field. The amount of shielding depends on the chemical composition of the molecule. This means that protons in different chemical compounds will be in slightly different field strengths and will therefore resonate at different frequencies. This change in resonant frequency from one compound to another is known as *chemical shift*. It can be used to perform chemical analysis in the technique of MR spectroscopy and to produce images based on chemical composition. However, in conventional MRI the chemical-shift effect can be the source of an unwanted artifact.

In tissue the chemical shift in resonant frequency between the fat and water is approximately 3.3 ppm, as shown in Figure 3-6. At a field strength of 1.5 T the protons have a basic resonant frequency of approximately 64 MHz. Multiplying this by 3.3 gives a water-fat chemical shift of approximately 210 Hz. At a field strength of 0.5 T the chemical shift would be only 70 Hz.

There are several imaging techniques that can be used to selectively image either the water or fat tissue components. One approach is to suppress either the fat or water signal with specially designed RF pulses. This technique is known as *spectral presaturation* and will be described in Chapter 8. Another technique makes use of the fact that the signals from water and fat are not always in step, or in phase, with each other and can be separated to create either water or fat images.

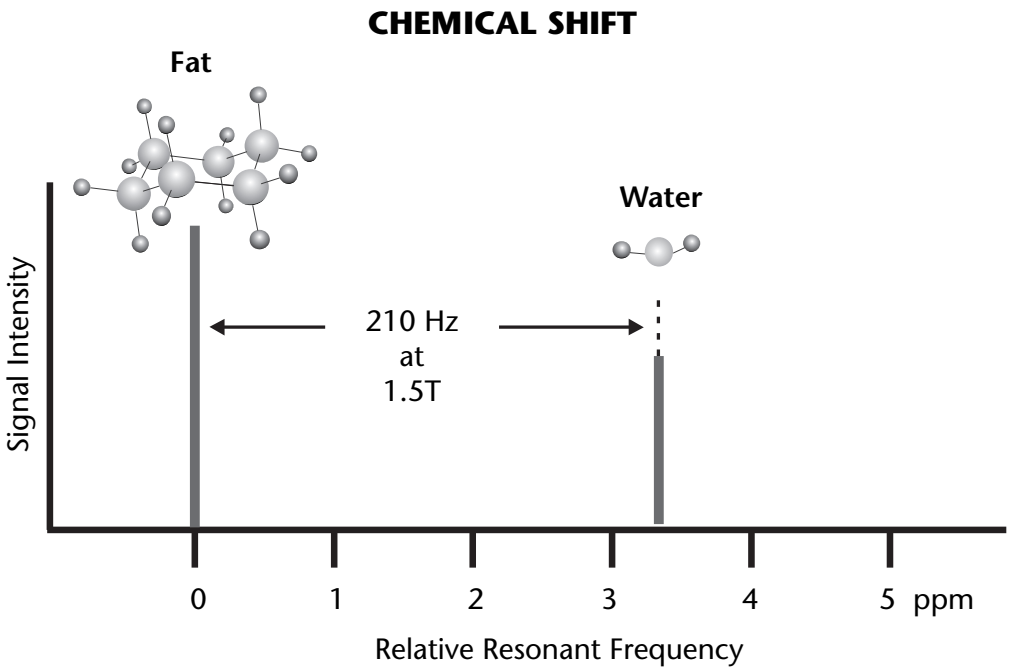
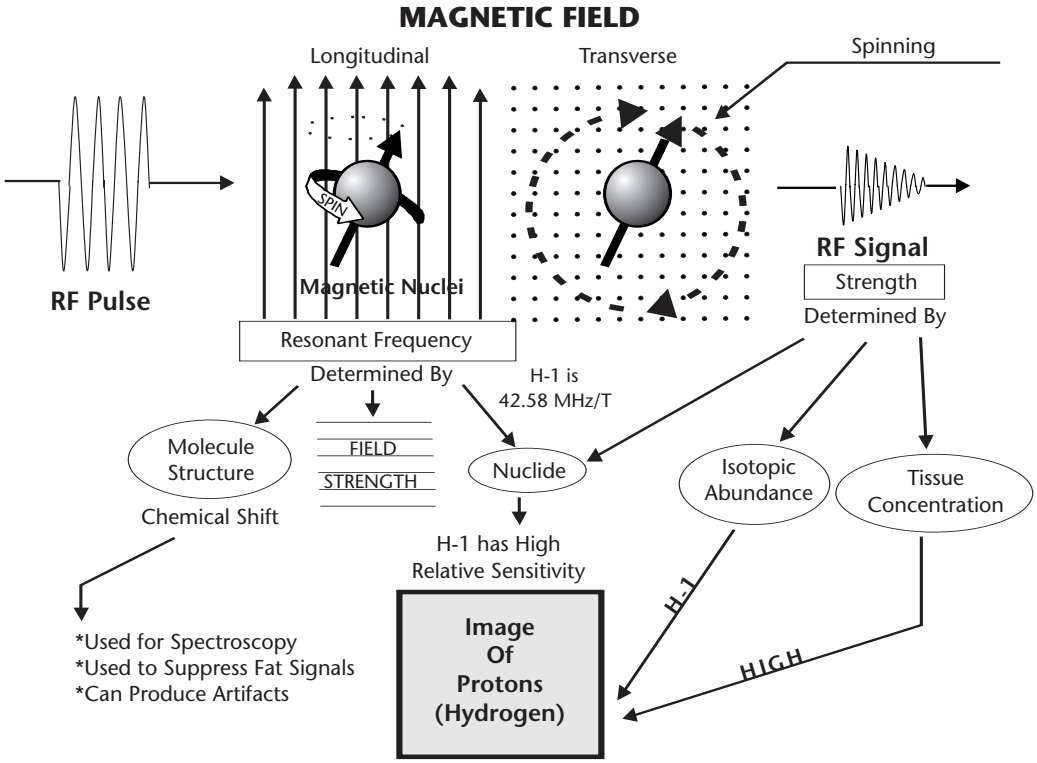


Figure 3-6. The chemical shift effect on the relative resonant frequency of protons in fat and in water.

Mind Map Summary
Nuclear Magnetic Resonance



When a magnetic nucleus is located in a strong magnetic field, it resonates. In effect, it becomes a tuned radio receiver and transmitter. The resonance occurs because the spinning nucleus precesses at a rate that is in the radio frequency range. The resonant frequency is determined by three factors. Each specific nuclide has a unique resonant frequency. The resonant frequency is affected to a small degree by the structure of the molecule containing the magnetic nucleus. This, the chemical shift effect, is useful for spectroscopy and to suppress fat signals in images. It can also lead to a certain type of image artifact. The resonant frequency is directly proportional to the strength of the magnetic field. This is useful because it makes it possible to tune the various parts of a body to different frequencies by applying magnetic field gradients.

When an RF pulse is applied to a magnetic nucleus oriented in the longitudinal direction, it can be flipped into the transverse plane. There the nucleus spins around the axis of the magnetic field and generates an RF signal. It is the signals from many spinning nuclei that are collected and used to form the image. It is necessary to have strong signals to produce good images. Signal strength depends on three factors. Each magnetic nuclide has a unique sensitivity or relative signal strength. All chemical elements have several different isotopes, but all isotopes of an element are usually not in the form of magnetic nuclei. Therefore, the abundance of the magnetic isotope

for a specific element has a major effect on signal strength. To produce strong signals a tissue must have a relatively high concentration of a chemical element and the most abundant isotope of that element must be magnetic.

Hydrogen is the only chemical element with a high concentration in tissue and body fluids in the form of an isotope that has a magnetic nucleus. Therefore, MR imaging is essentially limited to visualizing only one chemical element, hydrogen.

4

Tissue Magnetization And Relaxation

Introduction And Overview

We have considered the behavior of individual nuclei when placed in a magnetic field. MRI depends on the collective, or net, magnetic effect of a large number of nuclei within a specific voxel of tissue. If a voxel of tissue contains more nuclei aligned in one direction than in other directions, the tissue will be temporarily magnetized in that particular direction. This process is illustrated in Figure 4-1. In the absence of a magnetic field, the nuclei are randomly oriented and produce no net magnetic effect. This is the normal state of tissue before being placed in a magnetic field. When the tissue is placed in a magnetic field, and some of the nuclei align with the field, their combined effect is to magnetize the tissue in the direction of the magnetic field. A large arrow, the *magnetization vector*, is used to indicate the amount and direction of the magnetization. When tissue is placed in a magnetic field, the maximum magnetization that can be produced depends on three factors: (1) the concentration (density) of magnetic nuclei, typically protons, in the tissue voxel; (2) the magnetic sensitivity of the nuclide; and (3) the strength of the magnetic field. Since an imaging magnetic field aligns a very small fraction of the magnetic nuclei, the tissues are never fully magnetized. The amount of tissue magnetization determines the strength of the RF

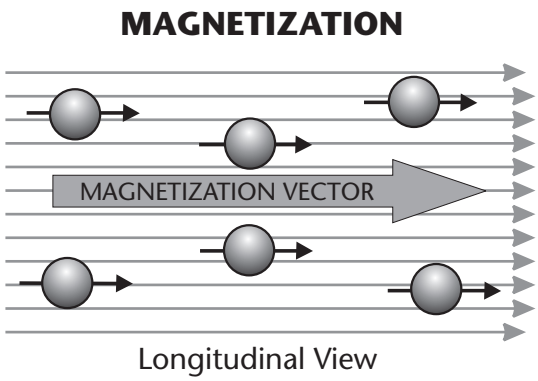


Figure 4-1. The magnetization of tissue produced by the alignment of magnetic nuclei (protons) in a magnetic field.

signals emitted by the tissue during an imaging or analytical procedure. This, in turn, affects image quality and imaging time, as explained in Chapter 10.

Let us recall that an MR image is an image of magnetized tissue and that the contrast we see is produced by different levels of magnetization that exist in the different tissues at the time when "the picture is snapped." As we will see in this chapter the level of magnetization at specific times during the imaging process is determined by the three tissue characteristics: proton density (PD), T1, and T2.

We will now see how these characteristics produce image contrast.

Tissue Magnetization

When tissue is placed in a magnetic field, it reaches its maximum magnetization within a few seconds and remains at that level unless it is disturbed by a change in the magnetic field or by pulses of RF energy applied at the resonant frequency. The MRI procedure is a dynamic process in which tissue is cycled through changes in its magnetization during each imaging cycle.

Magnetic Direction

The direction of tissue magnetization is specified in reference to the direction of the applied magnetic field, as shown in Figure 4-2. There are two principle directions that tissue is magnetized during the imaging process. Longitudinal magnetization is when the tissue is magnetized in a direction parallel to the direction of the field. Transverse magnetization is when the direction of tissue magnetization is at a 90° angle with respect to the direction of the magnetic field and is in the transverse plane.

Magnetic Flipping

The direction of tissue magnetization can be changed or flipped by applying a pulse of RF energy. This is done many times throughout the imaging process.

Flip Angle

The angle the magnetization is flipped is determined by the duration and strength of the RF pulse. Pulses are characterized by their *flip angles.*

Pulses with 90° and 180° flip angles are the most common but smaller flip angle pulses are also used in some imaging methods, such as gradient echo imaging.

The 90˚ Pulse, Saturation and Excitation

When a 90˚ pulse is applied to longitudinal magnetization, it flips it into the transverse plane as shown in Figure 4-3. This has two effects. First, it reduces the longitudinal magnetization to zero, a condition called *saturation.* It also produces transverse magnetization. As we will soon learn, transverse magnetization is an unstable or *excited* condition. Therefore, when a 90˚ pulse is applied to longitudinal magnetization, it produces both *saturation* of the longitudinal magnetization and a condition of *excitation* (transverse magnetization).

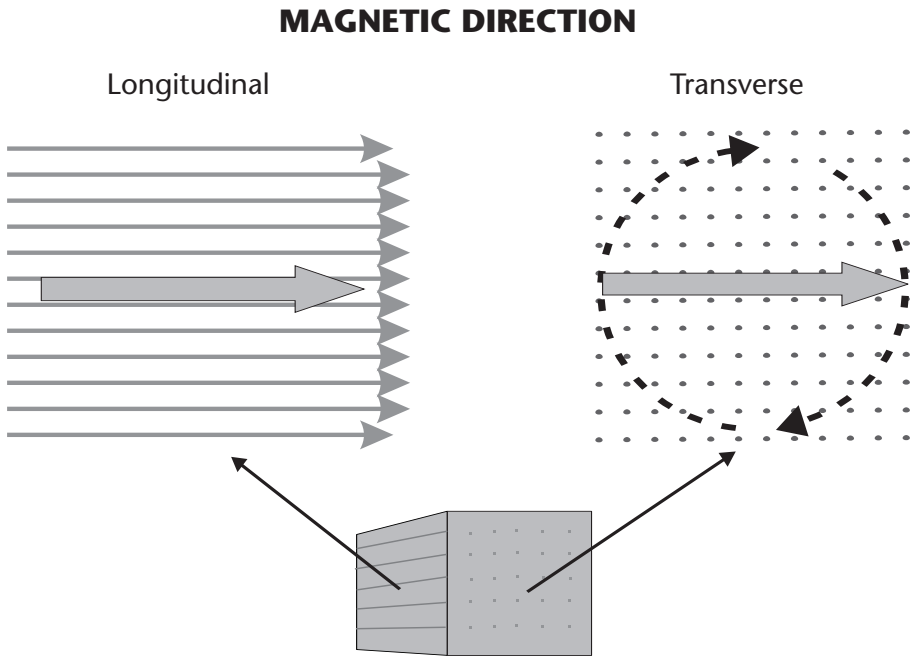


Figure 4-2. Longitudinal and transverse magnetization.

The actual direction of magnetization is not limited to longitudinal or transverse. It can exist in any direction. In principle, magnetization can have both longitudinal and transverse components. Since the two components have distinctly different characteristics, we consider them independently.

Longitudinal Magnetization And Relaxation

As we have seen, when tissue is placed in a magnetic field, it becomes magnetized in the longitudinal direction. It will remain in this state until the magnetic field is changed or until the magnetization is redirected by the application of an RF pulse. If the magnetization is temporarily redirected by an RF pulse, it will then, over a period of time, return to its original longitudinal position. If we consider only the longitudinal magnetization, it regrows after it has been reduced to zero, or saturated. This regrowth, or recovery, of longitudinal magnetization is the *relaxation* process, which occurs after saturation. The time required for the longitudinal magnetization to regrow, or relax, depends on characteristics of the material and the strength of the magnetic field.

Longitudinal magnetization does not grow at a constant rate, but at an exponential rate, as shown in Figure 4-4. An important concept to remember is that the MR image is an image of magnetized tissue with brightness indicating the level of magnetization. During the relaxation process, the level of magnetization is changing. Therefore, the brightness of tissue (if we could see it) is also changing as

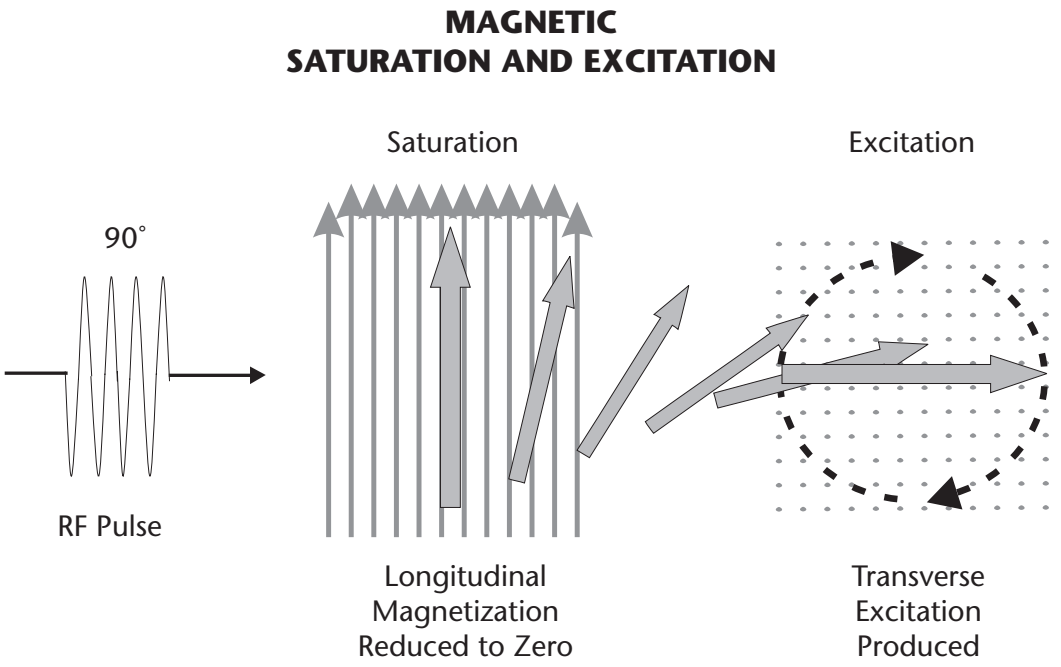


Figure 4-3. The application of a 90˚ RF pulse to longitudinal magnetization produces saturation of the longitudinal magnetization and creates transverse magnetization, an excited condition.

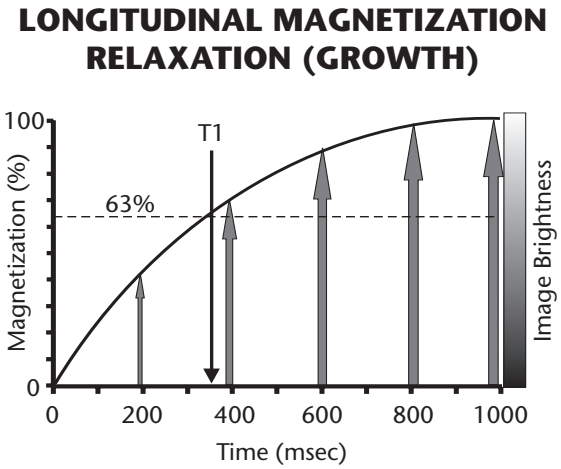


Figure 4-4. The growth of longitudinal magnetization (and tissue brightness) during the relaxation process following saturation.

indicated by the scale on the right of the illustration. Saturation turns the tissue dark and then it recovers brightness during the relaxation period.

The characteristic that varies from one type of tissue to another, and can be used to produce image contrast, is the time required for the magnetization to re-grow, or the relaxation time. Because of its exponential nature, it is difficult to determine exactly when the magnetization has reached its maximum. The convention is to specify the relaxation time in terms of the time required for the magnetization to reach 63% of its maximum. This time, the *longitudinal relaxation time*, is designated T1. The 63% value is used because of mathematical, rather than clinical, considerations. Longitudinal magnetization continues to grow with time, and reaches 87% of its maximum

after two T1 intervals, and 95% after three T1 intervals. For practical purposes, the magnetization can be considered fully recovered after approximately three times the T1 value of the specific tissue. We will see later that this must be taken into consideration when setting up an imaging procedure.

T1 Contrast

The time required for a specific level of longitudinal magnetization regrowth varies from tissue to tissue. Figure 4-5 shows the regrowth of two tissues with different T1 values. In this illustration we watch the intensity of brightness of a voxel of tissue during the relaxation process. Let us recall that the brightness of a tissue (RF signal intensity) is determined by the level of magnetization existing in a voxel of tissue at any instant in time. What we see in an image depends on when we "snap the picture" during the relaxation process. The important thing to notice is that the tissue with the shortest T1 has the highest level of magnetization at any particular time. The clinical significance of this is that tissues with short T1 values will be bright in T1-weighted images.

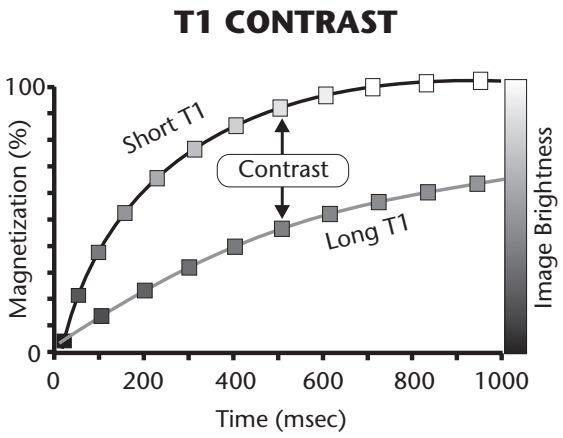


Figure 4-5. The formation of contrast between two tissues with different T1 values.

Table 4-1 lists typical T1 values for various tissues. Two materials establish the lower and upper values for the T1 range: fat has a short T1, and fluid falls at the other extreme (long T1). Therefore, in T1-weighted images, fat is generally bright, and fluid [cerebrospinal fluid (CSF), cyst, etc.] is dark. Most other body tissues are within the range between fat and fluid.

The longitudinal relaxation process involves an interaction between the protons and their immediate molecular environment. The rate of relaxation (T1 value) is related to the naturally occurring molecular motion. The molecular motion is determined by the physical state of the material and the size of the molecules. The relatively rigid structure of solids does not provide an environment for rapid relaxation, which results in long T1 values. Molecular motion in fluids, and fluid-like substances, is more inducible to the relaxation process. In this environment molecular size becomes an important characteristic.

Relaxation is enhanced by a general matching of the proton resonant frequency and the frequency associated with the molecular motions. Therefore, factors that change either of these two frequencies will generally have an effect on T1 values.

Molecular Size

Small molecules, such as water, have faster molecular motions than large molecules, such as lipids. The frequencies associated with the molecular motion of water molecules are both higher and more dispersed over a larger range for the larger molecules. This reduces the match between the frequencies of the protons and the frequencies of the molecular environment. This is why water and similar fluids have relatively long T1 values. Larger molecules, which have slower and less dispersed molecular movement, have a better frequency

Table 4-1. T2 and T1 Values for Various Tissues

Tissue	T2 (msec)	T1 (0.5 T) (msec)	T1 (1.5 T) (msec)
Adipose (Fat)	80	210	260
Liver	42	350	500
Muscle	45	550	870
White Matter	90	500	780
Gray Matter	100	650	920
CSF	160	1800	2400

match with the proton resonant frequencies. This enhances the relaxation process and produces short T1 values. Fat is an excellent example of a large molecular structure that exhibits this characteristic.

Tissues generally contain a combination of water and a variety of larger molecules. Some of the water can be in a relatively free state while other water is bound to some of the larger molecules. In general, the T1 value of the tissue is probably affected by the exchange of water between the free and the bound states. When the water is bound to larger molecular structures, it takes on the motion characteristics of the larger molecule. Factors such as a pathologic process, which alters the water composition of tissue, will generally alter the T1 values.

Magnetic Field Strength Effect

T1 values depend on the strength of the magnetic field. This is because the field strength affects the resonant frequency of the protons. As field strength is increased, the resonant frequency also increases and becomes less matched to the molecular motion frequencies. This results in an increase in T1 values, as indicated in Table 4-1.

Let us now combine two factors to create a T1 image as illustrated in Figure 4-6. One factor is that different tissues have different T1 values and rates of regrowth of longitudinal magnetization. This then causes the different tissues to be at different levels of magnetization (brightness) when the picture is snapped during the relaxation period. Here we see the order of tissue brightness is inversely related to T1 values. In principle, the tissues with short T1 values get brighter faster and are at a higher level when the picture is snapped.

Transverse Magnetization And Relaxation

Transverse magnetization is produced by applying a pulse of RF energy to the magnetized tissue. This is typically done with a 90° pulse, which converts longitudinal magnetization into transverse magnetization. Transverse magnetization is an unstable, or excited, condition and quickly decays after the termination of the excitation pulse. The decay of transverse magnetization is also a relaxation process, which can be characterized by specific relaxation times, or T2 values. Different types of tissue have different T2 values that can be used to discriminate among tissues and contribute to image contrast.

Transverse magnetization is used during the image formation process for two reasons: (1) to develop image contrast based on differences in T2 values; and (2) to generate the RF signals emitted by the tissue. Longitudinal

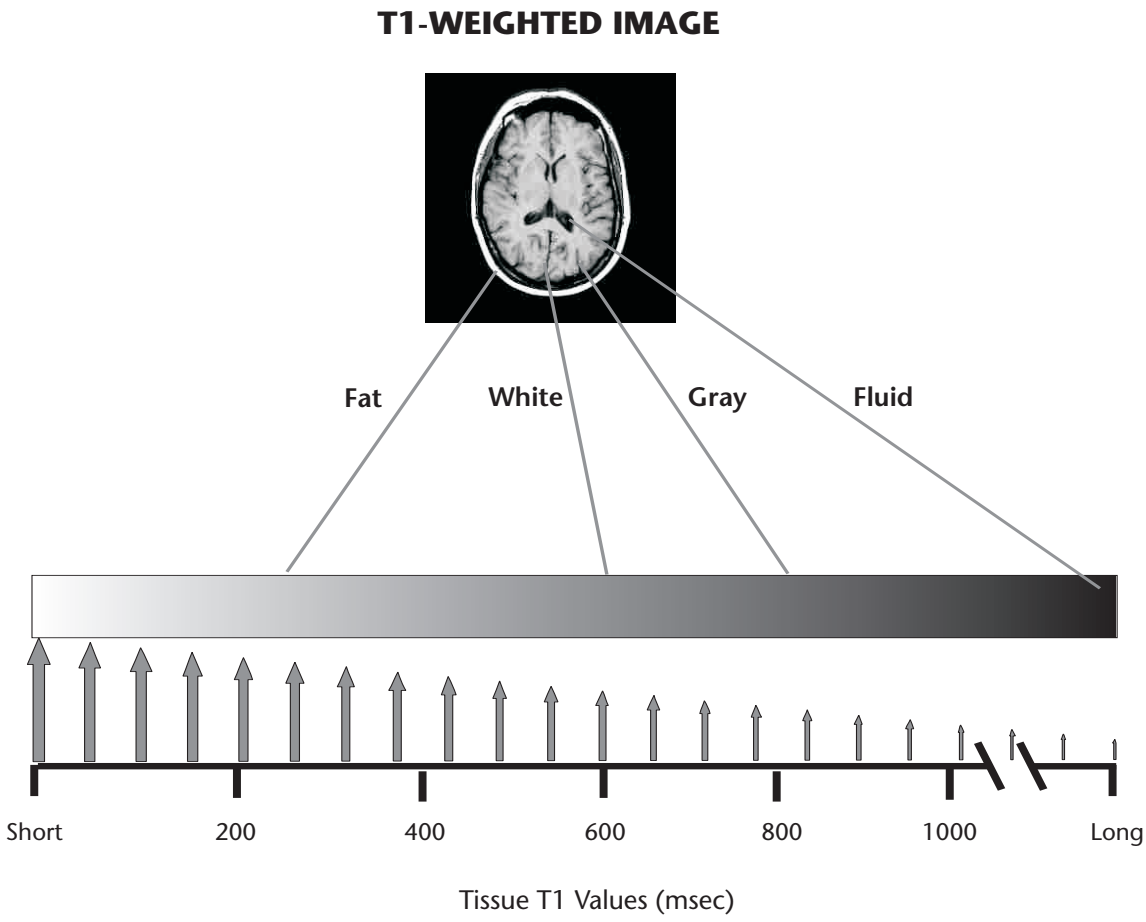


Figure 4-6. A T1 image showing the relationship of tissue brightness (signal intensity) to T1 values and level of magnetization during the longitudinal relaxation process.

magnetization is an RF silent condition and does not produce any signal. However, transverse magnetization is a spinning magnetic condition within each tissue voxel, and that generates an RF signal. As we will see in the next chapter, each imaging cycle must conclude with transverse magnetization to produce the RF signal used to form the image.

The characteristics of transverse magnetization and relaxation are quite different from those for the longitudinal direction. A major difference is that transverse magnetization is an unstable condition and the relaxation process results in the decay, or decrease, in magnetization, as shown in Figure 4-7. The T2 value is the time required for 63% of the initial magnetization to dissipate. After one T2, 37% of the initial magnetization is present.

T2 Contrast

The difference in T2 values of tissues is the source of contrast in T2-weighted images. This is illustrated in Figure 4-8. Here we watch two tissues, with different T2 values, during the relaxation process. We see that they are both getting darker with time as the magnetization

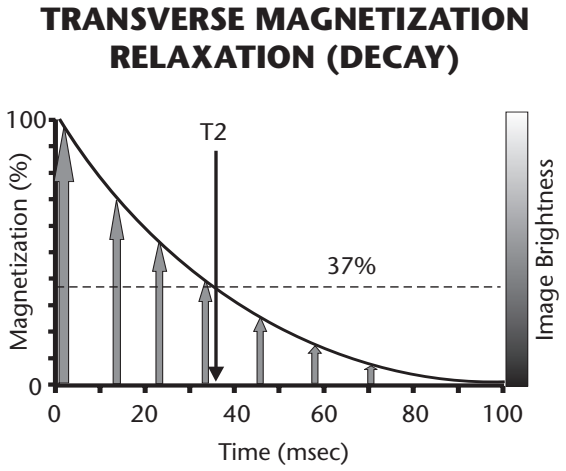


Figure 4-7. The decay of transverse magnetization during the relaxation process and the associated tissue brightness.

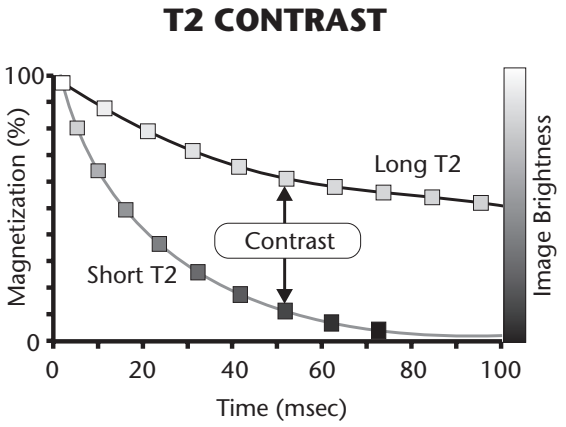


Figure 4-8. The formation of T2 contrast during the decay of transverse magnetization.

decays. However, they are not getting darker at the same rate. The tissue with the shorter T2 becomes darker faster leaving the tissue with the longer T2 to be bright at times during the relaxation time.

What we will actually see in a T2-weighted image, as shown in Figure 4-9, depends on the level of magnetization at the time when we snap the picture. The important thing to observe here is that the tissues with *long* T2 values are bright in T2 images.

In general, a T2-weighted image appears to be a reversal of a T1-weighted image. Tissues that are bright in one image are dark in the other image. This is because of a combination of two factors. One factor is that T1 and T2 values are generally related. Even though T2 values are much shorter than T1 values, as shown in Table 4-1, they are somewhat proportional. Tissues with long T1 values usually have long T2 values. The other factor is that the order of brightness in a T2 image is in the same direction as the T2 values. Remember, it was a reversed relationship for T1 images.

The decay of transverse magnetization (i.e., relaxation) occurs because of a dephasing among individual nuclei (protons) within the individual voxels, as shown in Figure 4-10.

Two basic conditions are required for transverse magnetization: (1) the magnetic moments of the nuclei must be oriented in the transverse direction, or plane; and (2) a majority of the magnetic moments must be in the same direction, or in phase, within the transverse plane. When a nucleus has a transverse orientation, it is actually spinning around an axis that is parallel to the magnetic field.

After the application of a 90˚ pulse, the nuclei have a transverse orientation and are rotating together, or in phase, around the magnetic field axis. This rotation or spin is a result of the normal precession discussed earlier. The precession rate, or resonant frequency, depends on the strength of the magnetic field where the nuclei are located. Nuclei located in field areas with different strengths spin (precess) at different rates. Even within a very small volume of tissue, nuclei are in slightly different magnetic field strengths. As a result, some nuclei spin faster than others. Also, there are interactions

(spin-spin interactions) among the spinning nuclei. After a short period of time, the nuclei are not spinning in phase. As the directions of the nuclei begin to spread and they dephase, the magnetization of the tissue decreases. A short time later, the nuclei are randomly oriented in the transverse plane, and there is no transverse magnetization.

Proton Dephasing

Two major effects contribute to the dephasing of the nuclei and the resulting transverse relaxation. In the imaging process the spin echo technique is used to separate the two sources of dephasing, as we will see in Chapter 6.

T2 Tissue Characteristics

One effect is the exchange of energy among the spinning nuclei (spin-spin interactions), which results in relatively slow dephasing and loss of magnetization. The rate at which this occurs is determined by characteristics of the tissue. It is this dephasing activity that is characterized by the T2 values as shown in Table 4-1.

T2* Magnetic Field Effects

A second effect, which produces relatively rapid dephasing of the nuclei and loss of transverse magnetization, is the inherent inhomogeneity of the magnetic field within

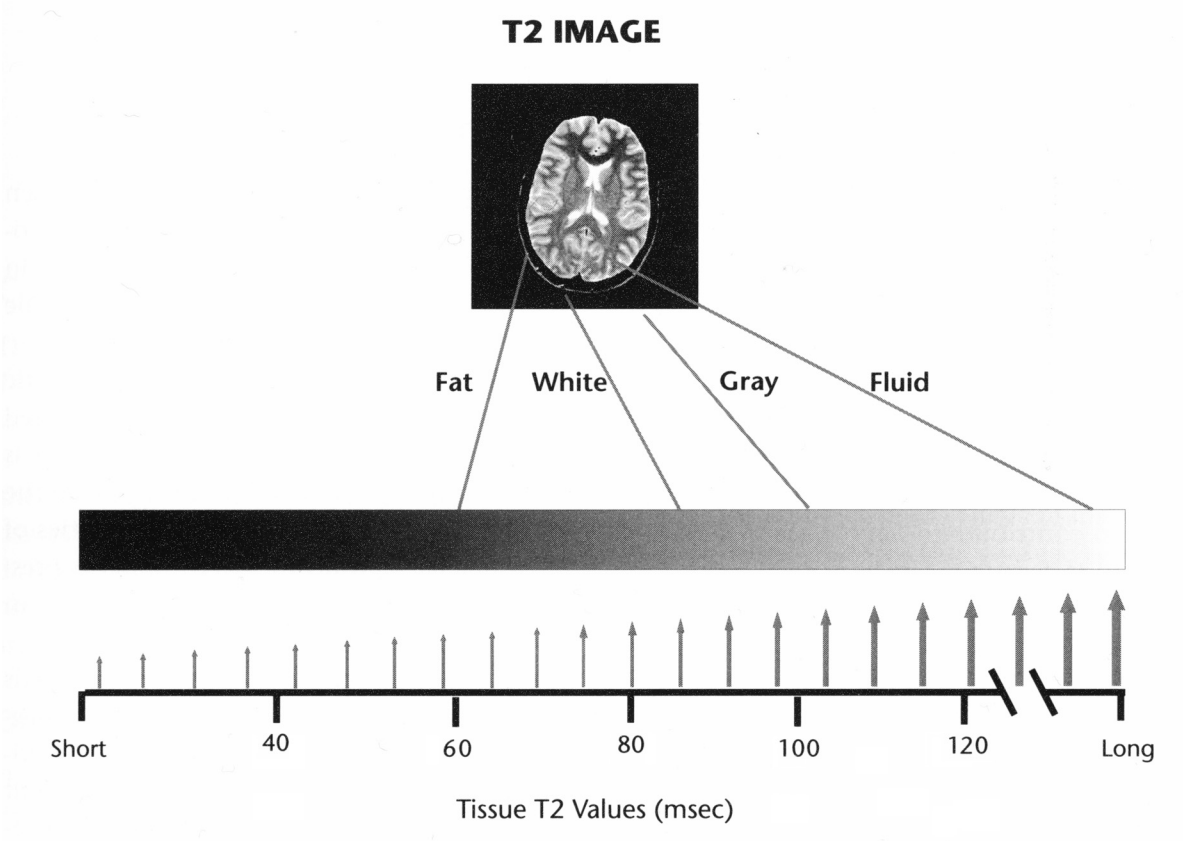


Figure 4-9. A T2 image showing the relationship of tissue brightness (signal intensity) to T2 values.

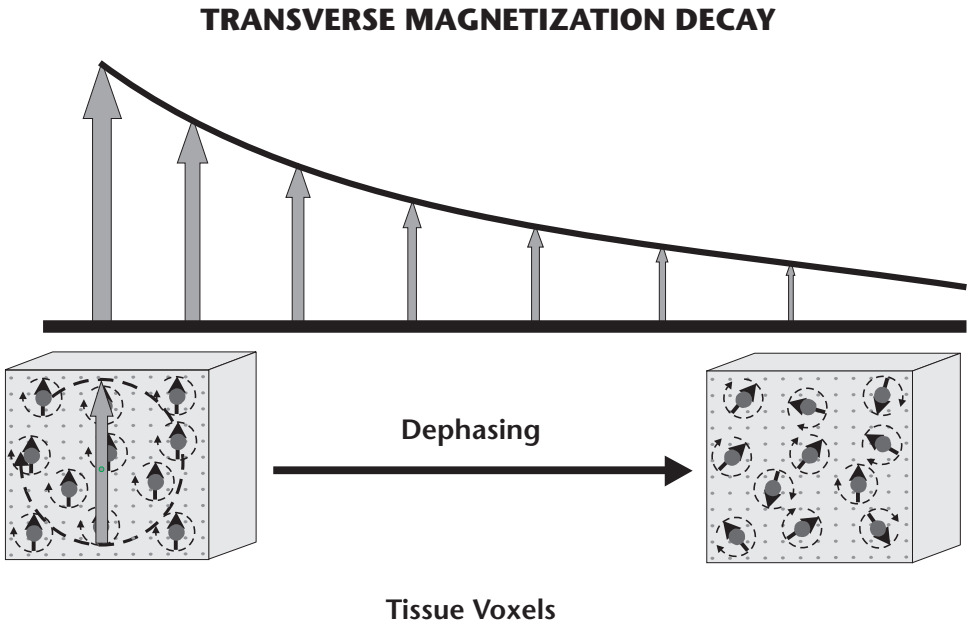


Figure 4-10. The dephasing of protons that produces transverse magnetization decay.

each individual voxel. The field inhomogeneities are sufficient to produce rapid dephasing. This effect, which is different from the basic T2 characteristics of the tissue, tends to mask the true relaxation characteristics of the tissue. In other words, the actual transverse magnetization relaxes much faster than the tissue characteristics would indicate. This real relaxation time is designated as T2*. The value of T2* is usually much less than the tissue T2 value, as illustrated in Figure 4-11. Several factors can contribute to field inhomogeneities and to T2* decay. One is the general condition of the magnetic field. Some fields are more homogeneous than others. Another factor is that different tissues or materials in the body might have different magnetic *susceptibilities*. *Susceptibility* is a characteristic of a material that determines its ability to become magnetized when it is in a magnetic field. If a region of tissue contains materials with different susceptibilities, this results in a reduction of field homogeneity.

Magnetic Susceptibility

The magnetization of tissue that we have been discussing is a *nuclear* magnetic effect produced by the alignment of magnetic nuclei in a magnetic field. Other materials can become magnetized by other, non-nuclear effects.

Many materials are *susceptible* to magnetic fields and become magnetized when located in fields. The susceptibility of a material is determined by the orbital electrons in the atom rather than the magnetic properties of the nucleus. Significant susceptibility is present only when there are unpaired electrons in the outer orbit.

There are three general types of materials with respect to magnetic susceptibility: *diamagnetic, paramagnetic*, and *ferromagnetic*. The primary characteristic of each type is the amount and direction of magnetization that the material develops when placed in a magnetic field. There are situations when each type plays a role in the MR imaging process.

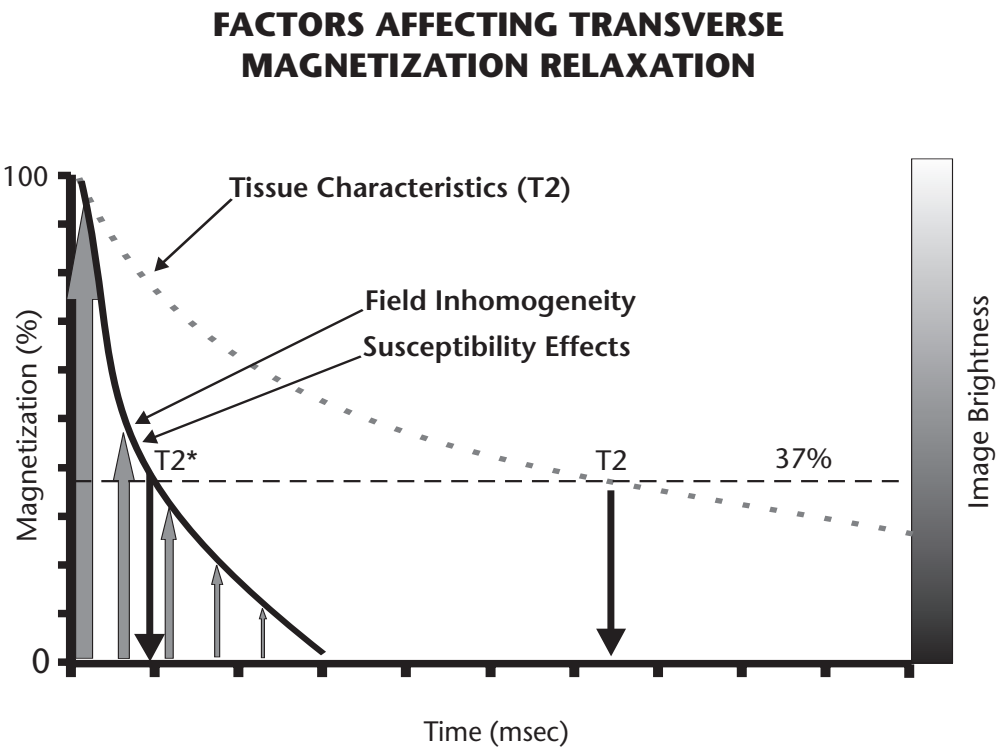


Figure 4-11. Comparison of relaxation produced by the T2 characteristics of tissue and the T2* effects associated with magnetic field inhomogeneities.

Contrast Agents

The inherent tissue characteristics (PD, T1, and T2) do not always produce adequate contrast for some clinical objectives. It is possible to administer materials (i.e., contrast agents) that will alter the magnetic characteristics within specific tissues or anatomical regions. There are several different types of contrast agents, which will now be considered. Contrast agents used in MRI are generally based on relaxation effects.

Diamagnetic Materials

Diamagnetic materials have negative and relatively low magnetic susceptibility. This means that they develop only low levels of magnetization and it is in a direction opposite to the direction of the magnetic field. Although many biological molecules are diamagnetic, this is not a significant factor in MR imaging.

Paramagnetic Materials

Paramagnetic materials play an important role in contrast enhancement. They are materials with unpaired electrons that give each atom a permanent magnetic property. In paramagnetic materials each atom is magnetically independent, which distinguishes it from other materials to be discussed later.

Paramagnetic substances include metal ions such as gadolinium, manganese, iron, and chromium. Other substances such as nitroxide free radicals and molecular oxygen also have paramagnetic properties.

Gadolinium has seven unpaired electrons in its orbit, which give it a very strong magnetic property. It must be chelated to reduce its toxicity. An example is gadolinium chelated to diethylene triamine penta-acetic acid (GaDTPA).

When a paramagnetic substance, such as gadolinium, enters an aqueous solution, it affects the relaxation rate of the existing protons. It does not produce a signal itself. In relatively low concentrations, the primary effect is to increase the rate of longitudinal relaxation and shorten the value of T1. In principle, the fluctuating magnetic field from the individual paramagnetic molecules enhances the relaxation rate. The primary result is an increase in signal intensity with T1-weighted images. It is classified as a positive contrast agent.

Signal intensity will generally increase with the concentration of the paramagnetic agents until a maximum intensity is reached. This intensity is very dependent on the imaging parameters. Higher concentrations will generally produce a reduction of signal intensity. This occurs because the transverse relaxation rate is also increased, which results in a shortening of the T2 value.

Superparamagnetic Materials

When materials with unpaired electrons are contained in a crystalline structure, they produce a stronger magnetic effect (susceptibility) in comparison with the independent molecules of a paramagnetic substance. The susceptibility of superparamagnetic materials is several orders of magnitude greater than that of paramagnetic materials. These materials are in the form of small particles. Iron oxide particles are an example.

The particles produce inhomogeneities in the magnetic field, which results in rapid dephasing of the protons in the transverse plane and a shortening of T2.

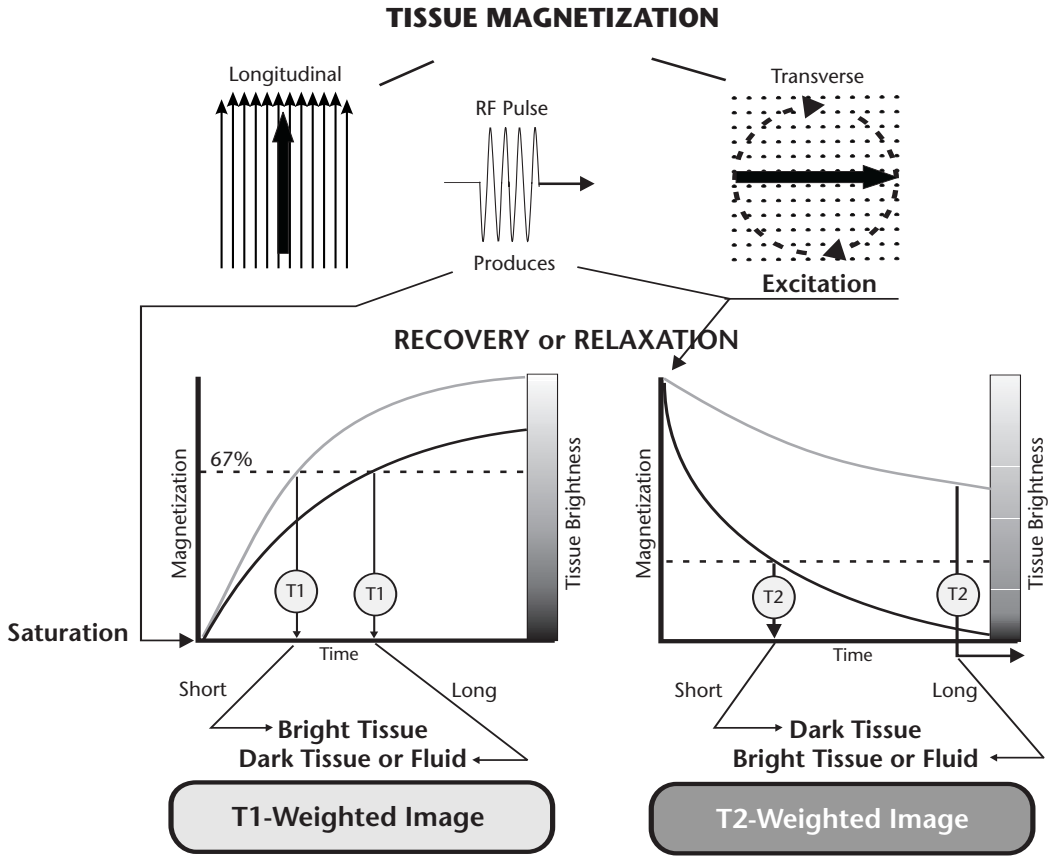
Superparamagnetic materials in the form of large particles generally reduce signal intensity and are classified as negative contrast agents. When in the form of very small particles, they reduce T1 and increase signal intensity.

Ferromagnetic Materials

Ferromagnetic is the name applied to iron and only a few other materials that have magnetic properties like iron. These materials have a very high susceptibility and develop a high level of magnetism when placed in a magnetic field.

Mind Map Summary

Tissue Magnetization And Relaxation



When tissue containing magnetic nuclei, i.e., protons, is placed in a strong magnetic field, the tissue becomes magnetized. It is initially magnetized in the longitudinal direction. However, by applying a pulse of RF energy the magnetization can be flipped into the transverse plane. Both longitudinal and transverse magnetization have characteristics that can be used to develop image contrast. An imaging procedure can be adjusted to display the different types of contrasts.

When a 90˚ RF pulse is applied to longitudinal magnetization, it produces two effects. First, it temporarily destroys the longitudinal magnetization, a condition known as *saturation*. It also produces transverse magnetization, a condition known as *excitation* because transverse magnetization is an unstable excited state.

After a saturation pulse is applied, the longitudinal magnetization will recover or regrow, a process known as *relaxation*. The rate of regrowth is a characteristic of each specific tissue and is described by its T1 value, the longitudinal relaxation time. A tissue with a short T1 will recover its

magnetization fast and will appear bright in a T1-weighted image. Tissues with longer T1 values will recover magnetization somewhat slower and will be relatively dark in T1-weighted images.

Following the production of transverse magnetization by the RF pulse the magnetization begins to decay or relax. The rate of relaxation is a characteristic of each specific tissue and is expressed by the T2 values, the transverse relaxation time. A tissue with a short T2 will lose its transverse magnetization rapidly and will appear relatively dark in T2-weighted images. Tissues and body fluids with long T2 values will retain their transverse magnetization longer and will appear bright in T2-weighted images.

5

The Imaging Process

Introduction And Overview

The MR imaging process consists of two major functions as shown in Figure 5-1. The first is the *acquisition* of RF signals from the patient's body and the second is the mathematical *reconstruction* of an image from the acquired signals.

In this chapter we will develop a general overview of the imaging process and set the stage for considering the different methods and techniques that are used to produce optimum images for various clinical needs.

k Space

During the acquisition process the signals are collected, digitized, and stored in computer memory in a configuration known as *k space*. The k space is divided into lines of data that are filled one at a time. One of the general requirements is that the k space must be completely filled before the image reconstruction can be completed. The size of k space (number of lines) is determined by the requirements for image detail and will be discussed in Chapters 9 thru 11.

Acquisition

The acquisition process consists of an imaging cycle that is repeated many times. The time required for a complete acquisition is determined by the duration of the cycle multiplied by the number of cycles. The duration of a cycle is TR (Time of Repetition), the adjustable protocol factor that is used to select the different

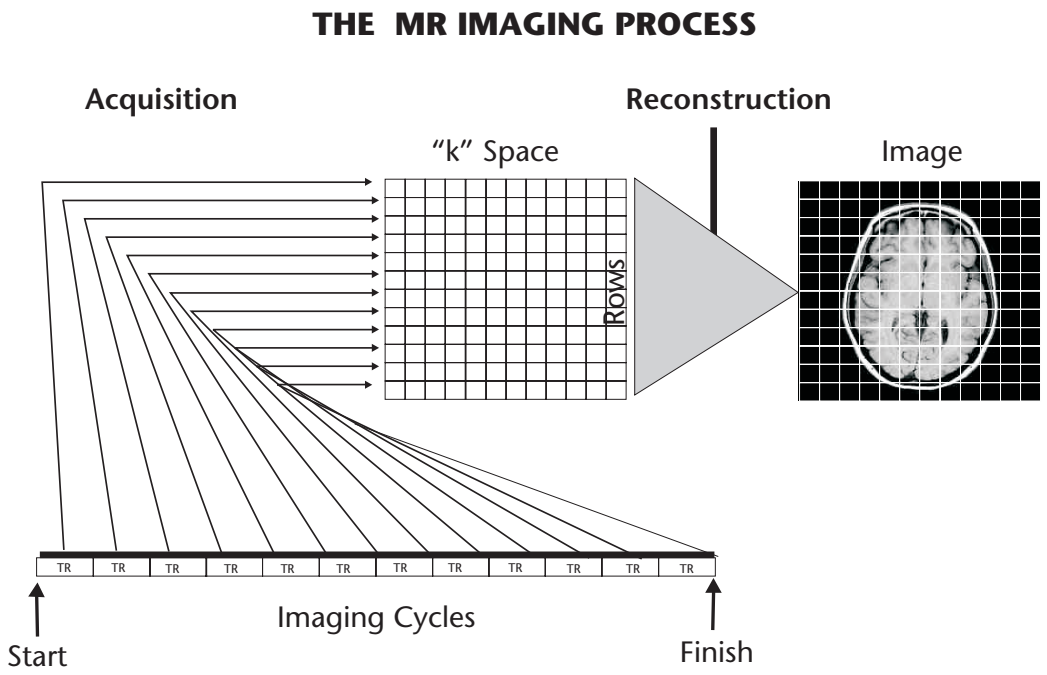


Figure 5-1. The two functions, *acquisition* and *reconstruction*, that make up the MR image production process.

types of image contrast. Also, the number of cycles used in an acquisition is adjustable. The number of cycles depends on the quality of the image that is required. The complete relationship between number of imaging cycles and image quality characteristics is described in Chapter 10.

Reconstruction

The image reconstruction process is usually fast compared to the acquisition process and generally does not require any decisions or adjustments by the operator.

Imaging Protocol

Each imaging procedure is controlled by a protocol that has been entered into the computer. Issues that must be considered in selecting, modifying, or developing a protocol for a specific clinical procedure include:

- The imaging method to be used
- The image types (PD, T1, T2, etc.)
- Spatial characteristics (slice thickness, number, etc.)
- Detail and visual noise requirements
- Use of selective signal suppression techniques
- Use of artifact reduction techniques

In the following chapters we will address each of these issues and the specific protocol factors that are used to produce the desired image characteristics.

Imaging Methods

There are several different imaging methods that can be used to create MR images. The principal difference among these methods is the sequence in which the RF pulses and gradients are applied during the acquisition process.

Therefore, the different methods are often referred to as the different *pulse sequences*. An overview of the most common methods is shown in Figure 5-2. As we see, the different methods are organized in a hierarchy structure. For each imaging method there is a set of factors that must be adjusted by the user to produce specific image characteristics.

The selection of a specific imaging method and factor values is generally based on requirements for contrast sensitivity to a specific tissue characteristic (PD, T1, T2) and acquisition speed. However, other characteristics such as visual noise and the sensitivity to specific artifacts might vary from method to method.

All of the imaging methods belong to one or both of the two major families, *spin echo* or *gradient echo*. The difference between the two families of methods is the process that is used to create the echo event at the end of each imaging cycle. For the spin echo methods, the echo event is produced by the application of a 180° RF pulse, as will be described in Chapter 6. For the gradient echo methods the event is produced by applying a magnetic field gradient, as described in Chapter 7. Each method has very specific characteristics and applications.

The Imaging Cycle

A common characteristic of all methods is that there are two distinct phases of the image acquisition cycle, as shown in Figure 5-3. One phase is associated with longitudinal magnetization and the other with transverse magnetization. In general, T1 contrast is developed during the longitudinal magnetization phase and T2 contrast is

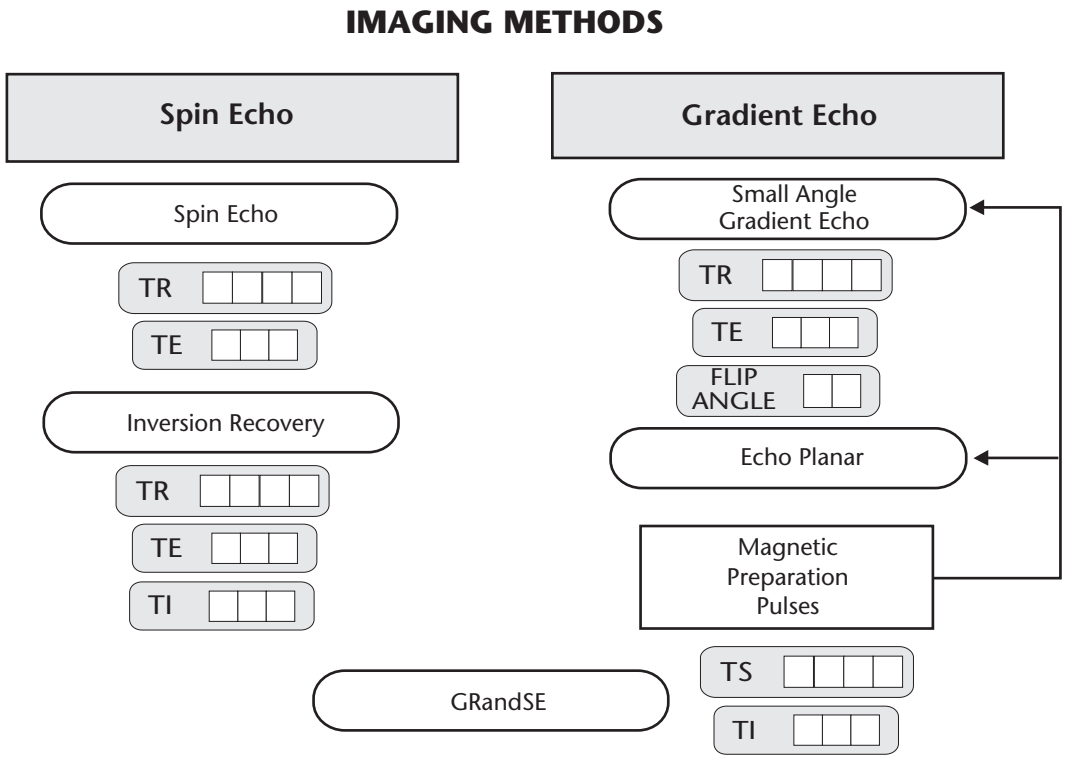


Figure 5-2. The principal spin echo and gradient echo imaging methods. GRandSE, or GRASE, is a combination of the two methods.

developed during the transverse magnetization phase. PD contrast is always present, but becomes most visible when it is not overshadowed by either T1 or T2 contrast. The predominant type of contrast that ultimately appears in the image is determined by the duration of the two phases and the transfer of contrast from the longitudinal phase to the transverse phase.

The duration of the two phases (longitudinal and transverse) is determined by the selected values of the protocol factors, TR (Time of Repetition) and TE (Time to Echo).

TR

TR is the time interval between the beginning of the longitudinal relaxation, following saturation, and the time at which the longitudinal magnetization is converted to transverse magnetization by the excitation pulse. This is when the picture is snapped relative to the longitudinal magnetization.

Because the longitudinal relaxation takes a relatively long time, TR is also the duration of the image acquisition cycle or the cycle repetition time (Time of Repetition).

TE

TE is the time interval between the beginning of transverse relaxation following excitation and when the magnetization is measured to produce image contrast. This happens at the *echo event* and is when the picture is snapped relative to the transverse magnetization. Therefore, TE is the Time to Echo event.

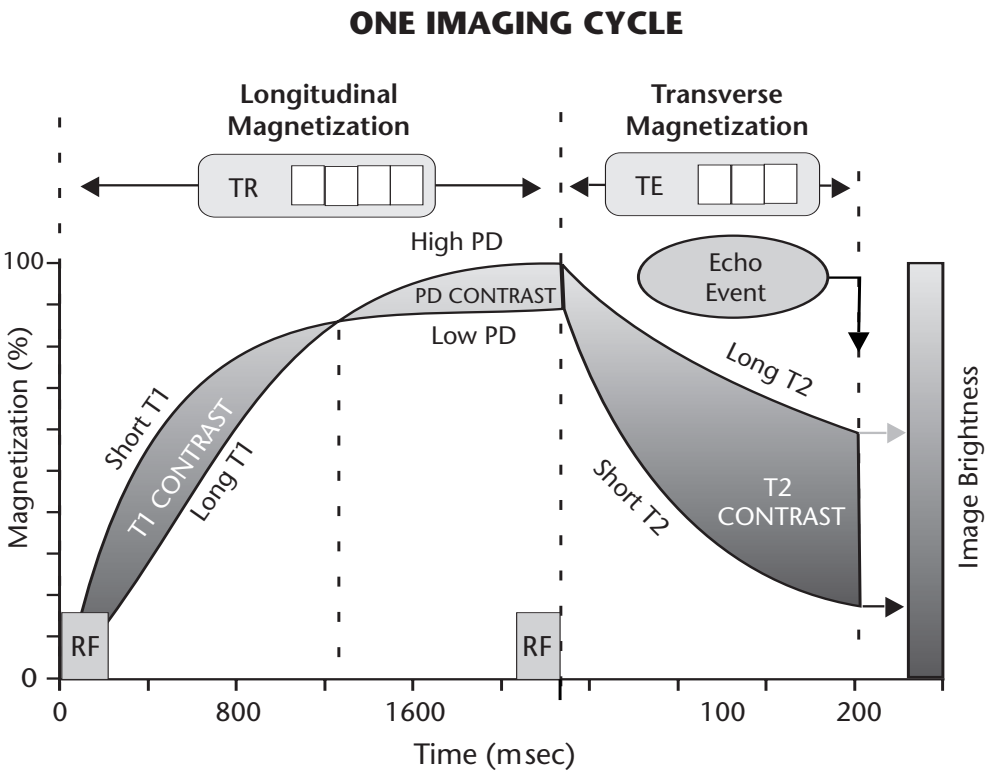


Figure 5-3. The longitudinal and transverse magnetization phases of an imaging cycle. T1 and PD contrast are produced during the longitudinal phase and T2 contrast is produced during the transverse phase.

Excitation

The transition from the longitudinal magnetization phase to the transverse magnetization phase is produced by applying an RF pulse. This is generally known as the excitation process because the transverse magnetization represents a more unstable or "excited" state than longitudinal magnetization.

The excitation pulse is characterized by a flip angle. A 90˚ excitation pulse converts all of the existing longitudinal magnetization into transverse magnetization. This type of pulse is used in the spin echo methods. However, there are methods that use excitation pulses with flip angles that are less than 90˚. Small flip angles (<90˚) convert only a fraction of the existing longitudinal magnetization into transverse magnetization and are used primarily to reduce acquisition time with the gradient echo methods described in Chapter 7.

The Echo Event and Signals

The transverse magnetization phase terminates with the echo event, which produces the RF signal. This is the signal that is emitted by the tissue and used to form the image. The echo event is produced by applying either an RF pulse or a gradient pulse to the tissue, as will be described in Chapters 6 and 7.

Contrast Sensitivity

In MRI the usual procedure is to select one of the tissue characteristics (PD, T1, T2) and then adjust the imaging process so that it has maximum, or at least adequate, contrast sensitivity for that specific characteristic. This produces an image that is heavily *weighted* by that characteristic. The contrast sensitivity of the imaging process and the resulting image contrast is determined by the specific imaging method and the combination of imaging protocol factor values, which we will consider in much more detail in later chapters. The discussion in this chapter will be based on the conventional spin echo method that uses only two factors, TR and TE, to control contrast sensitivity. However, it establishes some principles that apply to all methods.

T1 Contrast

During the relaxation (regrowth) of longitudinal magnetization, different tissues will have different levels of magnetization because of their different growth rates, or T1 values. Figure 5-4 compares two tissues with different T1 values.

The tissue with the shorter T1 value experiences a faster regrowth of longitudinal magnetization. Therefore, during this period of time it will have a higher level of magnetization, produce a more intense signal, and appear brighter in the image. In T1-weighted images brightness or high signal intensity is associated with short T1 values.

At the beginning of each imaging cycle, the longitudinal magnetization is reduced to zero (saturation) by an RF pulse, and then allowed to regrow, or relax. This is what happens in the spin echo method. In some other imaging methods, as we will see in the next two chapters, the cycle might begin with either partially saturated or inverted longitudinal magnetization. In all cases, T1 contrast is formed during the regrowth process. At a time determined by the selected TR value, the cycle is terminated and the magnetization value is converted to transverse, measured and displayed as a pixel intensity, or brightness, and a T1-weighted image is produced.

In principle, at the beginning of each imaging cycle all tissues are dark. As the tissues regain longitudinal magnetization, they become brighter. The brightness, or intensity, with which they appear in the image depends on

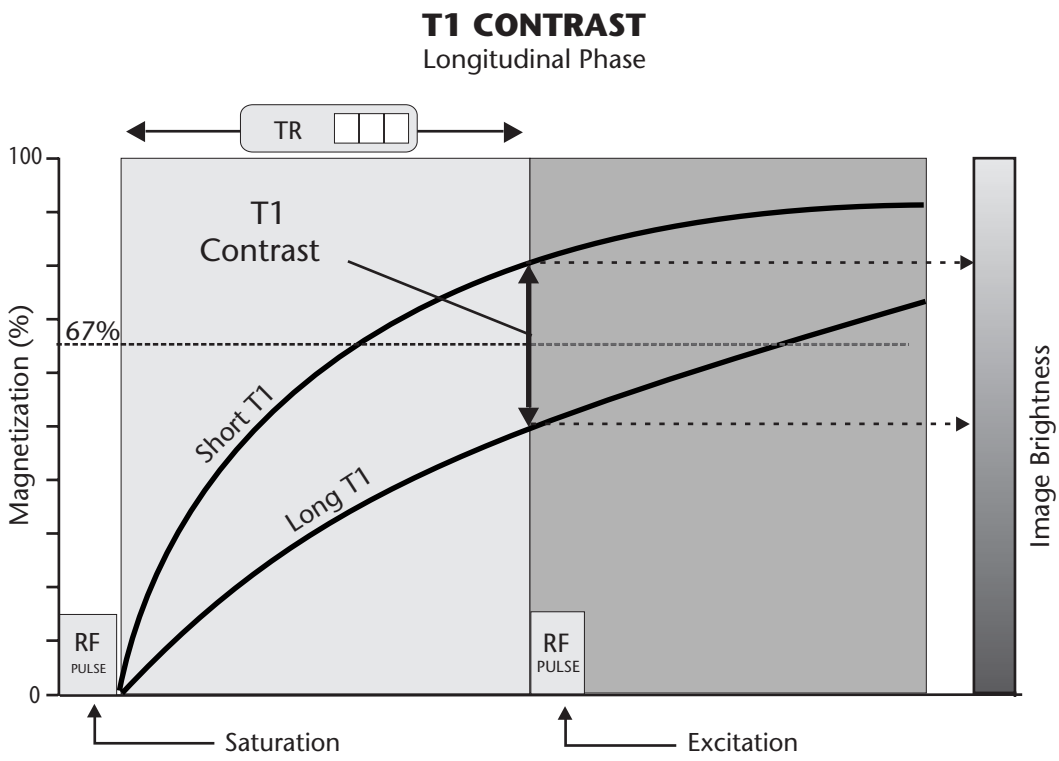


Figure 5-4. The amount of T1 contrast captured during the longitudinal magnetization phase is determined by the value of TR that is selected by the operator.

when during the regrowth process the cycle is terminated and the picture is snapped. This is determined by the selected TR value. When a short TR is used, the regrowth of the longitudinal magnetization is interrupted before it reaches its maximum. This reduces signal intensity and tissue brightness within the image but produces T1 contrast.

Increasing TR increases signal intensity and brightness up to the point at which magnetization is fully recovered, which is determined by the PD of each tissue. For practical purposes, this occurs when the TR exceeds approximately three times the T1 value for the specific tissues. Although it takes many cycles to form a complete image, the longitudinal magnetization is always measured at the same time in each cycle as determined by the setting of TR.

To produce a T1-weighted image, a value for TR must be selected to correspond with the time at which T1 contrast is significant between the two tissues. Several factors must be considered in selecting TR. If T1 contrast is represented by the ratio of the tissue magnetization levels, it is at its maximum very early in the relaxation process. However, the low magnetization levels present at that time do not generally produce adequate RF signal levels for many clinical applications. The selection of a longer TR produces greater signal strength but less T1 contrast.

The selection of TR must be appropriate for the T1 values of the tissues being imaged. If a TR value is selected that is equal to the T1 value of a tissue, the picture will be snapped when the tissue has regained 63% of its magnetization. This represents the time when there is

maximum contrast between tissues with small differences in T1 values.

Proton Density (PD) Contrast

The density, or concentration, of protons in each tissue voxel determines the maximum level of magnetization that can be obtained. Differences in PD among tissues can be used to produce image contrast, as illustrated in Figure 5-5. Here we see the growth of longitudinal magnetization for two tissues with the same T1 values but different relative PDs. The tissue with the lowest PD (80) reaches a maximum magnetization level that is only 80% that of the other tissue. The difference in magnetization levels at any point in time is because of the difference in PD and is therefore the source of PD contrast.

Although there is some PD contrast early in the cycle, it is generally quite small in comparison to the T1 contrast.

The basic difference between T1 contrast and PD contrast is that T1 contrast is produced by the rate of growth (relaxation), and PD contrast is produced by the maximum level to which the magnetization grows. In general, T1 contrast predominates in the early part of the relaxation phase, and PD contrast predominates in the later portion. T1 contrast gradually gives way to PD contrast as magnetization approaches the maximum value. A PD-weighted image is produced by selecting a relatively long

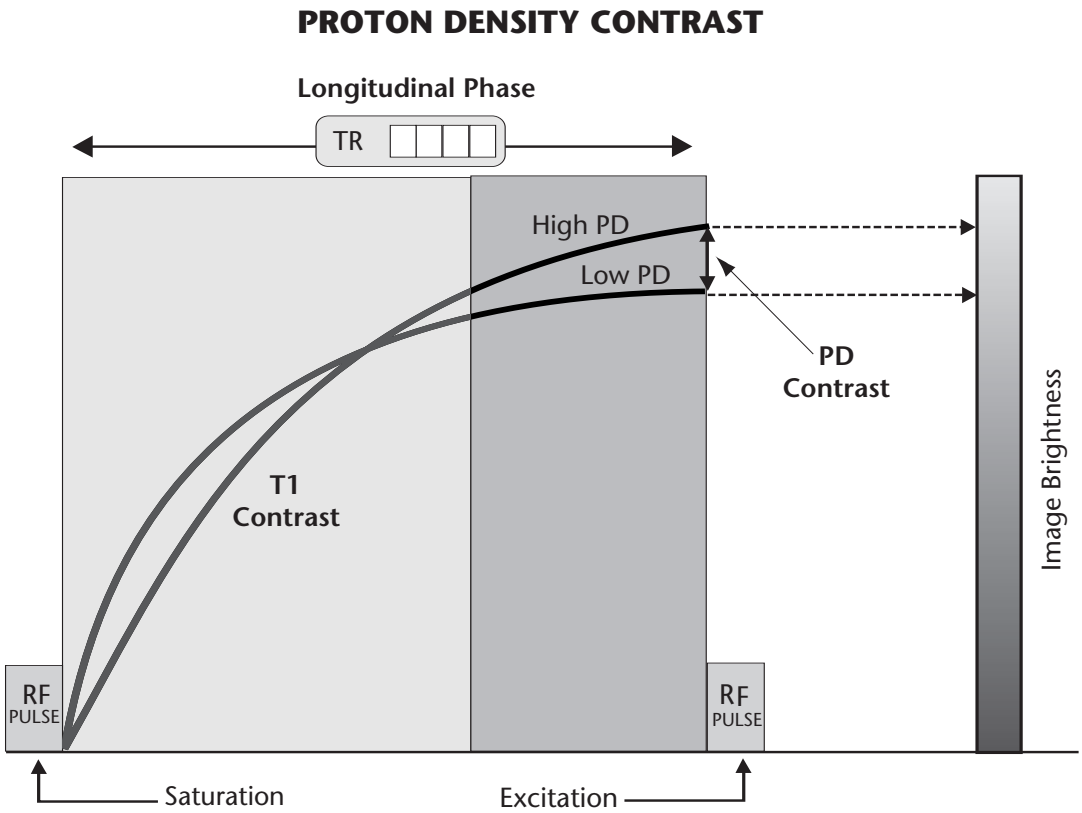


Figure 5-5. Proton density (PD) contrast is captured by setting TR to relatively long values. At that time the magnetization is determined by PD and is not T1 as in the earlier part of the cycle.

TR value so that the image is created or "the picture is snapped" in the later portion of the relaxation phase, where tissue magnetizations approach their maximum values. The TR values at which this occurs depend on the T1 values of the tissues being imaged.

It was shown earlier that tissue reaches 95% of its magnetization in three T1s. Therefore, a TR value that is at least three times the T1 values for the tissues being imaged produces almost pure PD contrast.

T2 Contrast

Now let us turn our attention to the transverse phase. During the decay of transverse magnetization, different tissues will have different levels of magnetization because of different decay rates, or T2 values. As shown in Figure 5-6, tissue with a relatively long T2 value will have a higher level of magnetization, produce a more intense signal, and appear brighter in the image than a tissue with a shorter T2 value.

Figure 5-6 shows the decay of transverse magnetization for tissues with different T2 values. The tissue with the shortest T2 value loses its magnetization faster than the other tissues.

The difference in T2 values of tissue can be translated into image contrast. For the purpose of this illustration we assume that the two tissues begin their transverse relaxation with the levels of magnetization determined by the PD. This is the usual case where the PD contrast

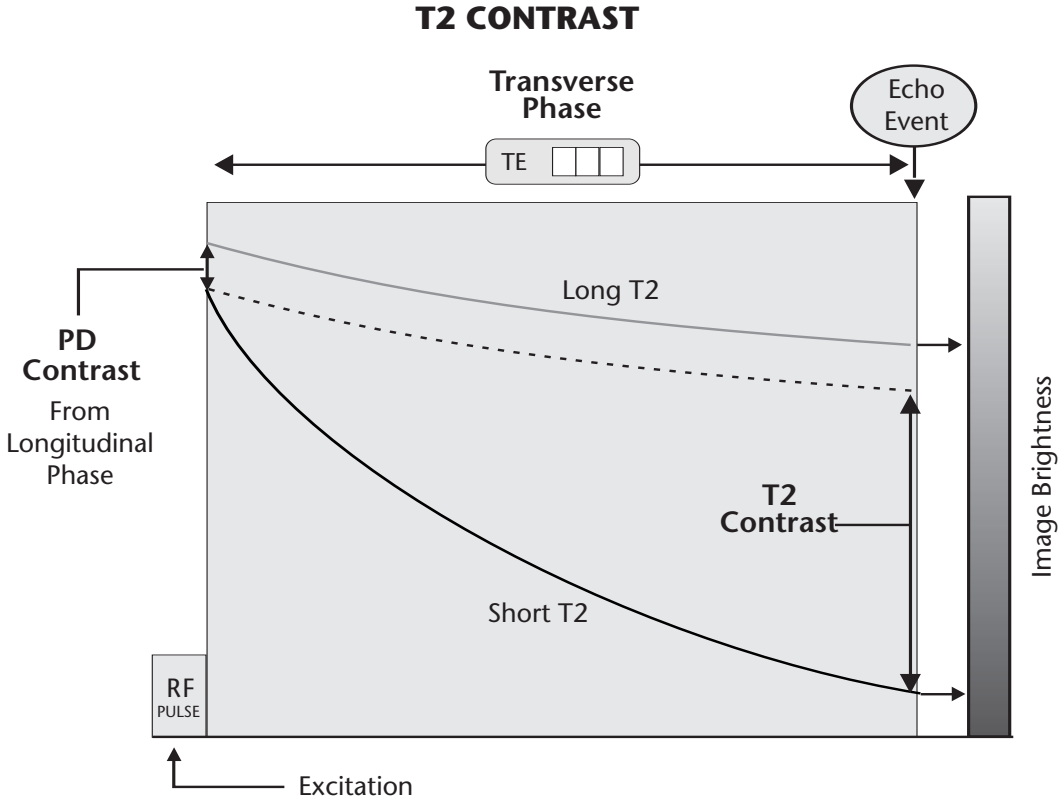


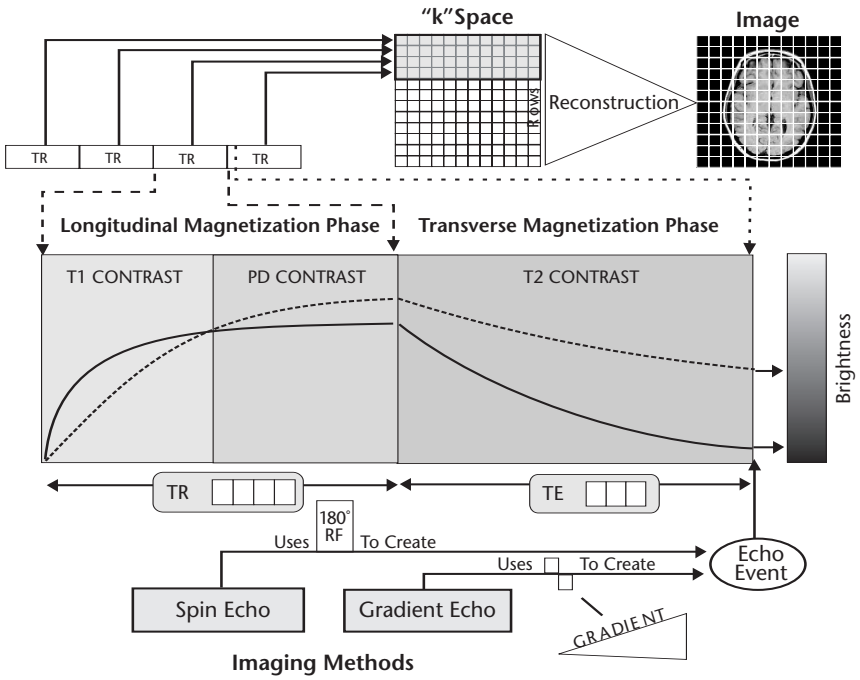
Figure 5-6. The formation of T2 contrast during the transverse magnetization phase. The amount of T2 contrast captured depends on the selected value of TE, the Time to Echo event.

present at the end of the longitudinal phase carries over to the beginning of the transverse phase. In effect, the transverse phase begins with PD contrast but adds T2 contrast as time elapses. The decay of the magnetization proceeds at different rates because of the different T2 values. The tissue with the longer T2 value maintains a higher level of magnetization than the other tissue and will remain bright longer. The difference in the tissue magnetizations at any point in time represents contrast.

At the beginning of the cycle there is no T2 contrast, but it develops and increases throughout the relaxation process. At the echo event the magnetization levels are converted into RF signals that are displayed as image pixel brightness; this is the time to echo event (TE) and is selected by the operator. Maximum T2 contrast is generally obtained by using a relatively long TE. However, when a very long TE value is used, the magnetization and the RF signals might be too low to form a useful image. In selecting TE values, a compromise must often be made between T2 contrast and good signal intensity.

The transverse magnetization characteristics of tissue (T2 values) are, in principle, added to the longitudinal characteristics carried over from the longitudinal phase (e.g., T1 and PD) to form the MR image. Usually we do not want to add T2 contrast to T1 contrast. That is because these two types of contrast oppose each other. Remember in Chapter 1 we saw that tissues that are bright in T1 images are dark in T2 images. This means that if we were to mix T1 and T2 contrast in the same image, one would cancel the other. When setting up a protocol for a T2 image it is necessary to use a long TR (in addition to a long TE) so that no, or very little, T1 contrast carries over to the transverse phase.

Mind Map Summary
The Imaging Process



The MR imaging process is one of creating contrast among tissues based on their magnetic characteristics. The primary characteristics are proton density (PD), T1, and T2. It is a dynamic activity in which the magnetization levels of the various tissues are undergoing almost constant change. During each imaging cycle there are two distinct magnetization phases: longitudinal and transverse. Different types of contrast are developed in each of these phases.

After application of a saturation pulse, which reduces the longitudinal magnetization to zero, the magnetization begins to regrow, a process known as relaxation. The rate of regrowth for a specific tissue is determined by that tissue's T1 value. Tissues with short T1 values grow faster than tissues with long T1 values. During this regrowth, T1 contrast in the form of different levels of magnetization is created among the tissues. This is the contrast that will be displayed in an image if the protocol parameters are set to produce a T1-weighted image. In a T1-weighted image, tissues with short T1 values will be bright. Tissues and fluid with long T1 values will be darker.

When an RF pulse is applied to longitudinal magnetization, it converts (flips) it to transverse magnetization, an unstable excited magnetic condition that decays with time. This decay process is the transverse magnetization relaxation process. The rate of decay of a specific tissue depends on that tissue's T2 value. Tissues with short T2 values decay faster than tissues with longer T2 values. When the imaging protocol factors are set to produce a T2-weighted image, tissues with short T2 values will be dark and tissues and fluids with longer T2 values will be bright.

6

Spin Echo Imaging Methods

Introduction And Overview

Spin echo is the name of the process that uses an RF pulse to produce the echo event. It is also the name for one of the specific imaging methods within the spin echo family of imaging methods; all of which use the spin echo process. We will first discuss the spin echo process and see how an RF pulse can produce an echo event and signal and then consider the spin echo methods.

The Spin Echo Process

The decay of transverse magnetization (i.e., relaxation) occurs because of dephasing among individual nuclei, as described in Chapter 4.

Let us recall that two basic conditions are required for transverse magnetization: (1) the magnetic moments of the nuclei must be oriented in the transverse direction, or plane, and (2) a majority of the moments must be in the same direction within the transverse plane. When a nucleus has a transverse orientation, it is actually precessing or rotating around an axis that is parallel to the magnetic field.

After the application of a 90˚ excitation pulse, the nuclei have a transverse orientation and are precessing together, or in-phase, around the magnetic field axis. This is the normal precession discussed earlier but flipped into the transverse plane. However, within an individual voxel some nuclei precess or spin faster than others. After a short period of time, the nuclei

are not spinning in-phase. As the directions of the nuclei begin to spread, the magnetization of the tissue decreases. A short time later, the nuclei are randomly oriented in the transverse plane; there is no transverse magnetization.

The two factors that contribute to the dephasing of the nuclei and the resulting transverse relaxation will now be reviewed again here. One is an exchange among the spinning nuclei (spin-spin interactions), which results in relatively slow dephasing and loss of magnetization. The rate at which this occurs is determined by characteristics of the tissue. It is this dephasing activity that is characterized by the T2 values and the source of contrast that we want to capture in T2 images. A second factor, which produces relatively rapid dephasing of the nuclei and loss of transverse magnetization, is the inhomogeneity of the magnetic field. Even within a small volume of tissue, the field inhomogeneities are sufficient to produce rapid dephasing. This effect, which is generally unrelated to the T2 characteristics of the tissue, tends to mask the true relaxation characteristics of the tissue. In other words, the actual transverse magnetization relaxes much faster than the tissue characteristics would indicate. We remember that this real relaxation time is designated as T2*. The value of T2* is always much less than the tissue T2 value. As a result, the transverse magnetization disappears before T2 contrast can be formed.

We are about to discover that spin echo is a process for recovering the lost transverse magnetization and making it possible to produce images of the three tissue characteristics, including T2.

An RF signal is produced whenever there is transverse magnetization. Immediately after an excitation pulse, a so-called free induction decay (FID) signal is produced. The intensity of this signal is proportional to the level of transverse magnetization. Both decay rather rapidly because of the magnetic field inhomogeneities just described. The FID signal is not used in the spin echo methods. It is used in the gradient echo methods to be described in Chapter 7.

The spin echo process is used to compensate for the dephasing and rapid relaxation caused by the field inhomogeneities and to restore the magnetization to the level that depends only on the tissue T2 characteristics. The sequence of events in the spin echo process is illustrated in Figure 6-1.

Transverse magnetization is produced with a 90˚ RF excitation pulse that flips the longitudinal magnetization into the transverse plane. Immediately following the RF pulse, each voxel is magnetized in the transverse direction. However, because of the local magnetic field inhomogeneities within each voxel, the protons precess at different rates and quickly slip out of phase. This produces the rapid decay characterized by T2* and the associated FID signal. At this time the protons are still rotating in the transverse plane, but they are out of phase.

If a 180˚ pulse is applied to the tissue containing these protons, it flips the protons around an axis in the transverse plane; this reverses their direction of rotation as illustrated in Figure 6-2. This causes the fast protons to be located behind the slower ones. As the faster protons begin to catch up with the slower ones, they regain a common alignment, or come back into phase. This, in turn, causes the transverse magnetization to reappear and form the echo event. However, the magnetization does not grow to the initial value because the relaxation (dephasing) produced by the tissue is not reversible. The rephasing of the protons causes the magnetization to build up to a level determined by the T2 characteristics of the tissue. As soon as the magnetization reaches this

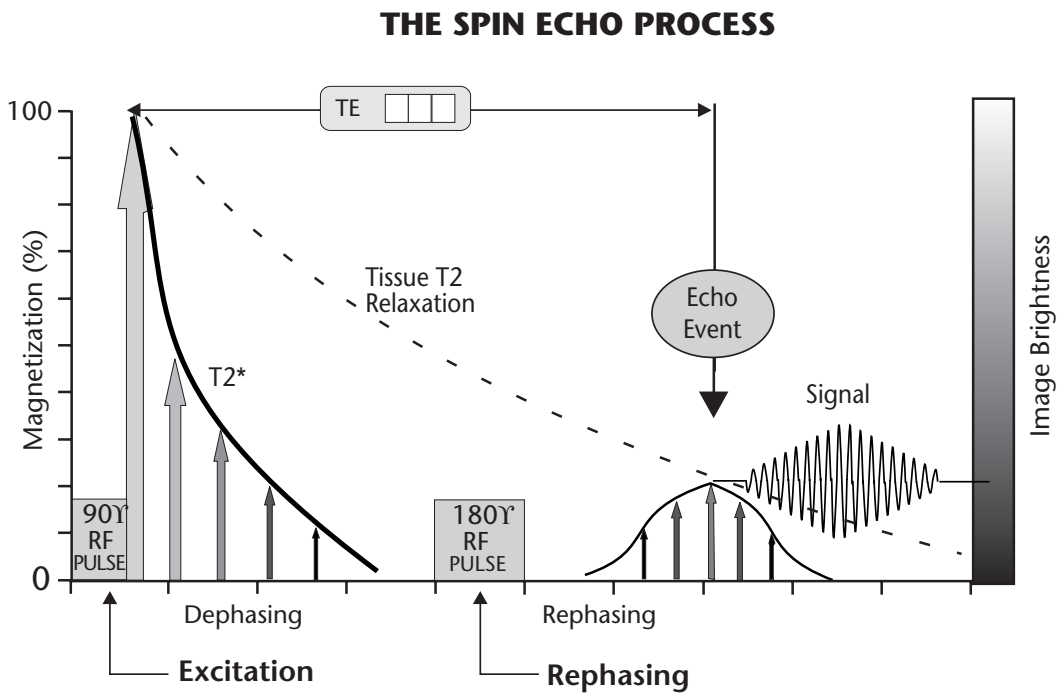


Figure 6-1. The spin echo process showing the use of a 180˚ pulse to rephase the protons and to produce an echo event.

maximum, the protons begin to move out of phase again, and the transverse magnetization dissipates. Another 180˚ pulse can be used to produce another rephasing. In fact, this is what is done in multi-echo imaging and will be described later in this chapter.

RF Pulse Sequence

The different imaging methods are produced by the type (flip angle) and time intervals between the applied RF pulses. The basic pulse sequence for the spin echo method is shown in Figure 6-3. Each cycle begins with a 90˚ excitation pulse that produces the initial transverse magnetization and a later 180˚ pulse that rephases the protons to produce the echo event.

The time between the initial excitation and the echo signal is TE. This is controlled by adjusting the time interval between the 90˚ and the 180˚ pulses, which is 1/2 TE.

The Spin Echo Method

This method can be used to produce images of the three basic tissue characteristics: PD, T1, and T2. The sensitivity to a specific characteristic is determined by the values selected for the two time intervals or imaging factors, TR and TE.

The process of creating images with the three types of contrast (PD, T1, and T2) described in the last chapter was a description of the spin echo method. There we saw that the type of image that was produced depended on the values selected for the two protocol factors, TR and TE. We will now review that process with a few more details specifically as it applies to the spin echo method.

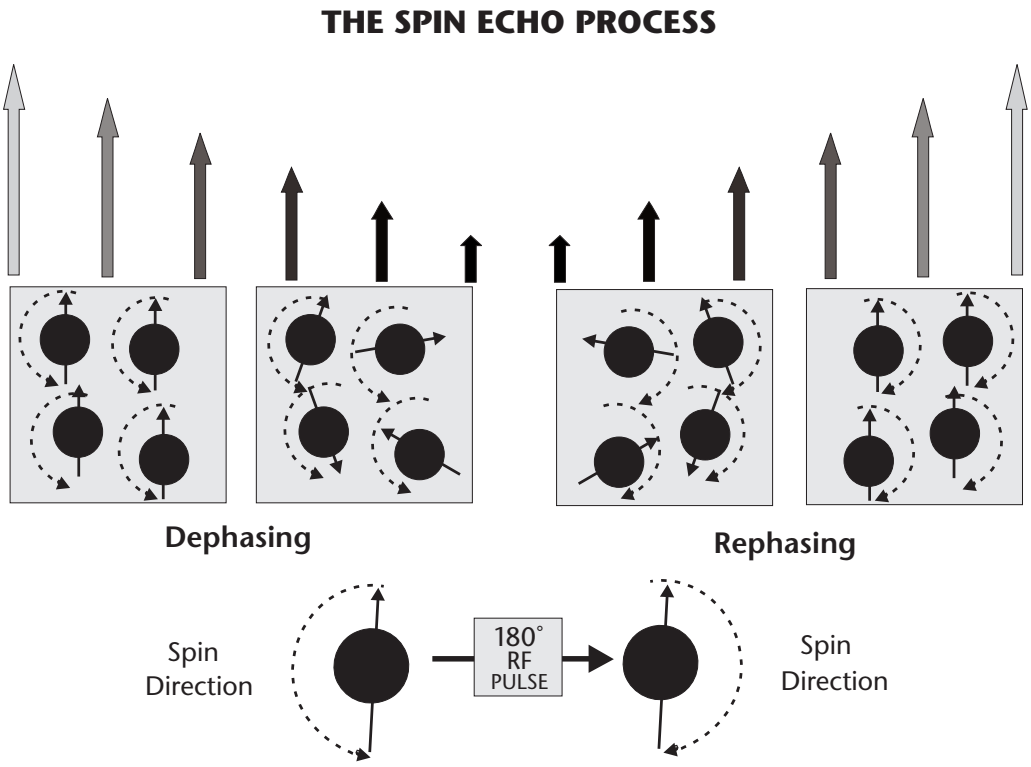


Figure 6-2. The 180° pulse sets up the protons so that they rephase.

Proton Density (PD) Contrast

PD contrast develops as the longitudinal magnetization approaches its maximum, which is determined by the PD of each specific tissue. Therefore, relatively long TR values are required to produce a PD-weighted image. Short TE values are generally used to reduce T2 contrast contamination and to maintain a relatively high signal intensity.

T1 Contrast

To produce image contrast based on T1 differences between tissues, two factors must be considered. Since T1 contrast develops during the early growth phase of longitudinal magnetization, relatively short TR values must be used to capture the contrast. The second factor is to preserve the T1 contrast during the time of transverse relaxation. The basic problem is that if T2 contrast is allowed to develop, it generally counteracts T1 contrast. This is because tissues with short T1 values usually have short T2 values. The problem arises because tissues with short T1s are generally bright, whereas tissues with short T2s have reduced brightness when T2 contrast is present. T2 contrast develops during the TE time interval. Therefore, a T1-weighted image is produced by using short TR values and short TE values.

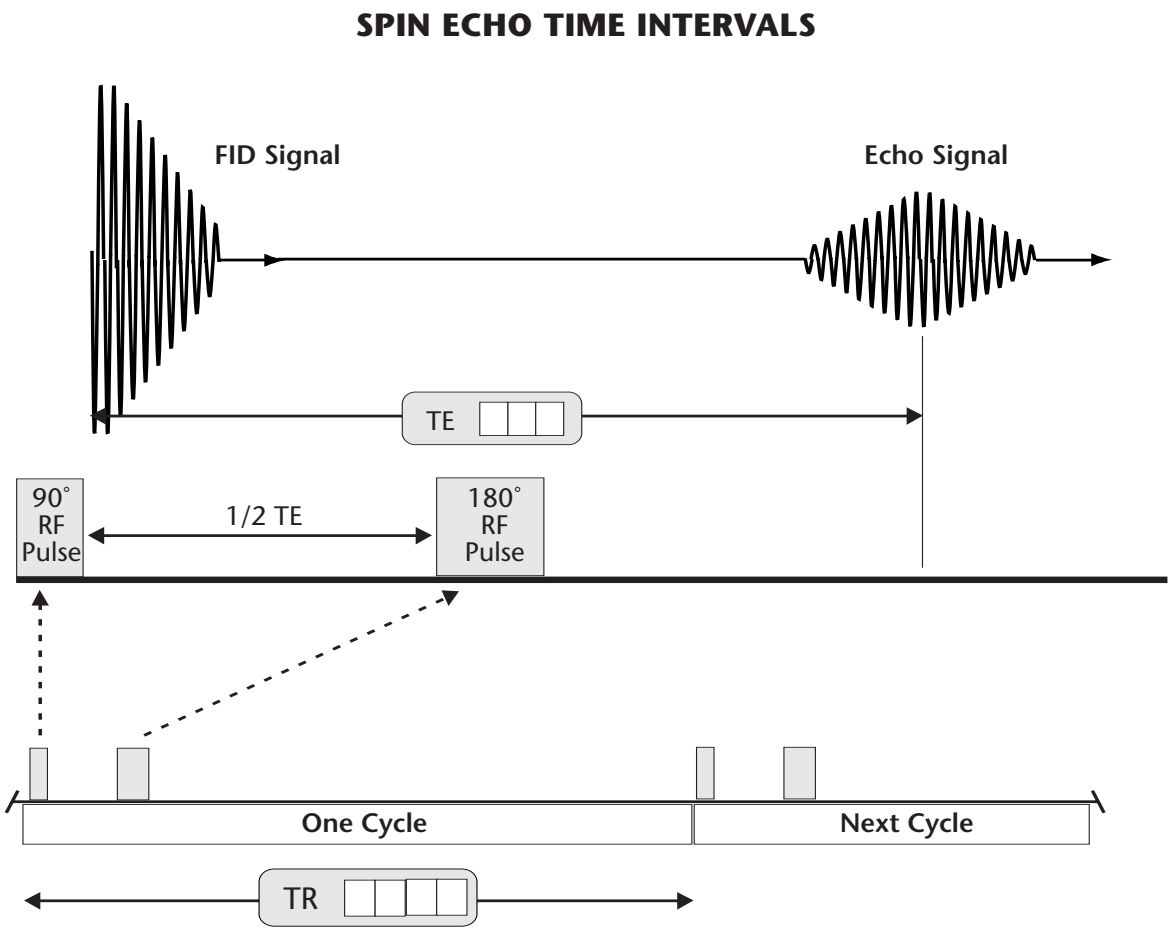


Figure 6-3. The RF pulses and time intervals in a spin echo imaging cycle.

T2 Contrast

The first step in producing an image with significant T2 contrast is to select a relatively long TR value. This minimizes T1 contrast contamination and the transverse relaxation process begins at a relatively high level of magnetization. Long TE values are then used to allow T2 contrast time to develop.

The spin echo method is the only method that produces true T2 contrast. That is because it is able to rephase the protons and remove the T2* effect.

Multiple Spin Echo

It is possible to produce a series of echo events within one cycle as illustrated in Figure 6-4. This is done by applying several 180˚ pulses after each 90˚ excitation pulse. The advantage is that echo events with different TE values are produced in one acquisition cycle. Separate images are formed for each TE value. This makes it possible to create both a PD image (short TE) and a T2 image (long T2) in the same acquisition.

Table 6-1 summarizes the combination of TR and TE values used to produce the three

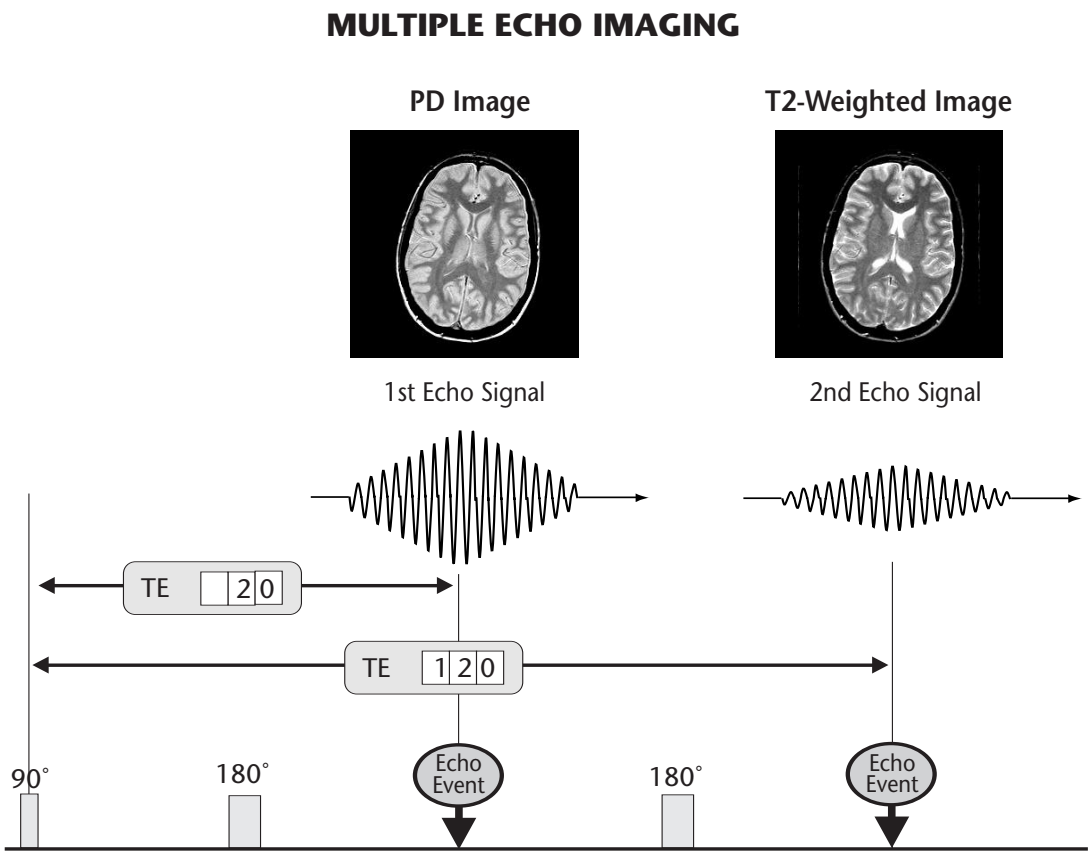


Figure 6-4. A multiple spin echo imaging that produces both a PD and T2 image in the same acquisition.

basic image types with the spin echo method. Optimum values of TR and TE for a specific protocol might vary because of considerations for other factors such as image acquisition time, number of slices, etc.

Inversion Recovery

Inversion recovery is a spin echo imaging method used for several specific purposes. One application is to produce a high level of T1 contrast and a second application is to suppress the signals and resulting brightness of fat and fluids. The inversion recovery pulse sequence is obtained by adding an additional 180˚ pulse to the conventional spin echo sequence, as shown in Figure 6-5. The pulse is added at the beginning of each cycle where it is applied to the longitudinal magnetization carried over from the previous cycle. Each cycle begins as the 180˚ pulse inverts the direction of the longitudinal magnetization. The regrowth (recovery) of the magnetization starts from a negative (inverted) value, rather than from zero, as in the spin echo method.

The inversion recovery method, like the spin echo method, uses a 90˚ excitation pulse to produce transverse magnetization and a final 180˚ pulse to produce a spin echo signal. That is

Table 6-1. Selection of TR and TE values to produce the three image types with spin echo method. Values shown are typical but can be varied to some extent to accommodate specific imaging conditions.

	T1 Image	PD Image	T2 Image
TR	Short (500 msec)	Long (2000 msec)	Long (2000 msec)
TE	Short (15-20 msec)	Short (15-20 msec)	Long (120 msec)

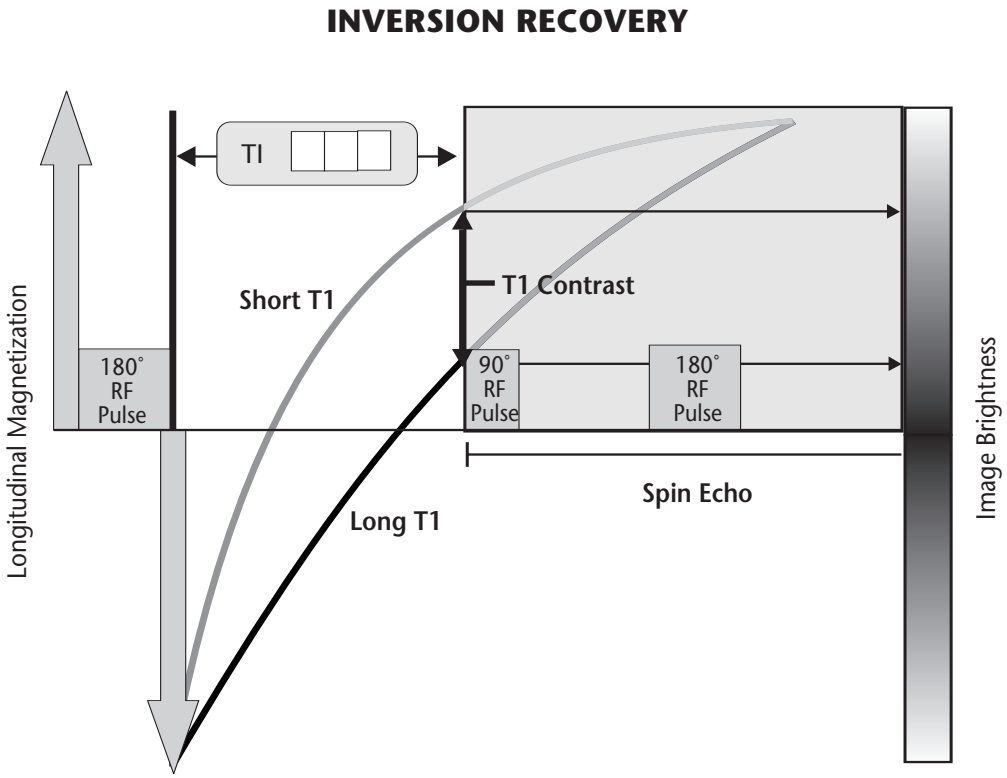


Figure 6-5. The inversion recovery method with TI set to produce an image with high T1 contrast.

why it is classified as one of the spin echo, rather than gradient echo, methods. An additional time interval is associated with the inversion recovery pulse sequence. The time between the initial 180˚ pulse and the 90˚ pulse is designated the **T**ime after **I**nversion (TI). It can be varied by the operator and used as a contrast control.

T1 Contrast

The principal characteristic of many inversion recovery images is high T1 contrast. This occurs because the total longitudinal relaxation time is increased because it starts from the inverted state. There is more time for the T1 contrast to

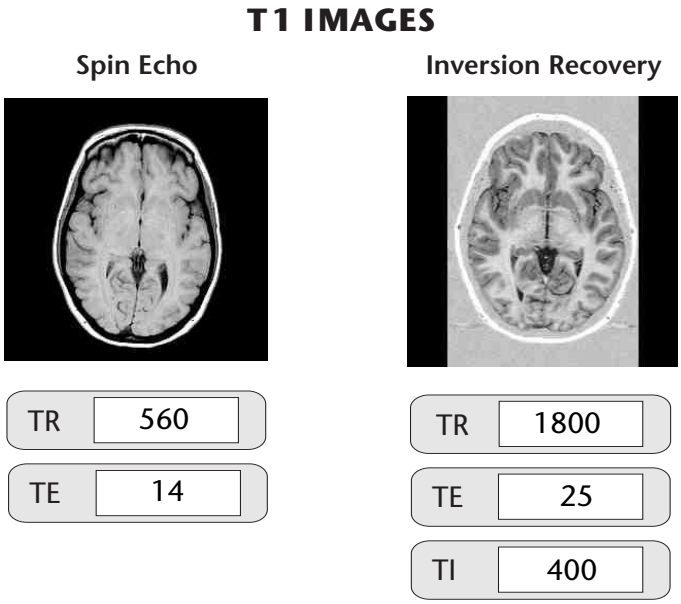
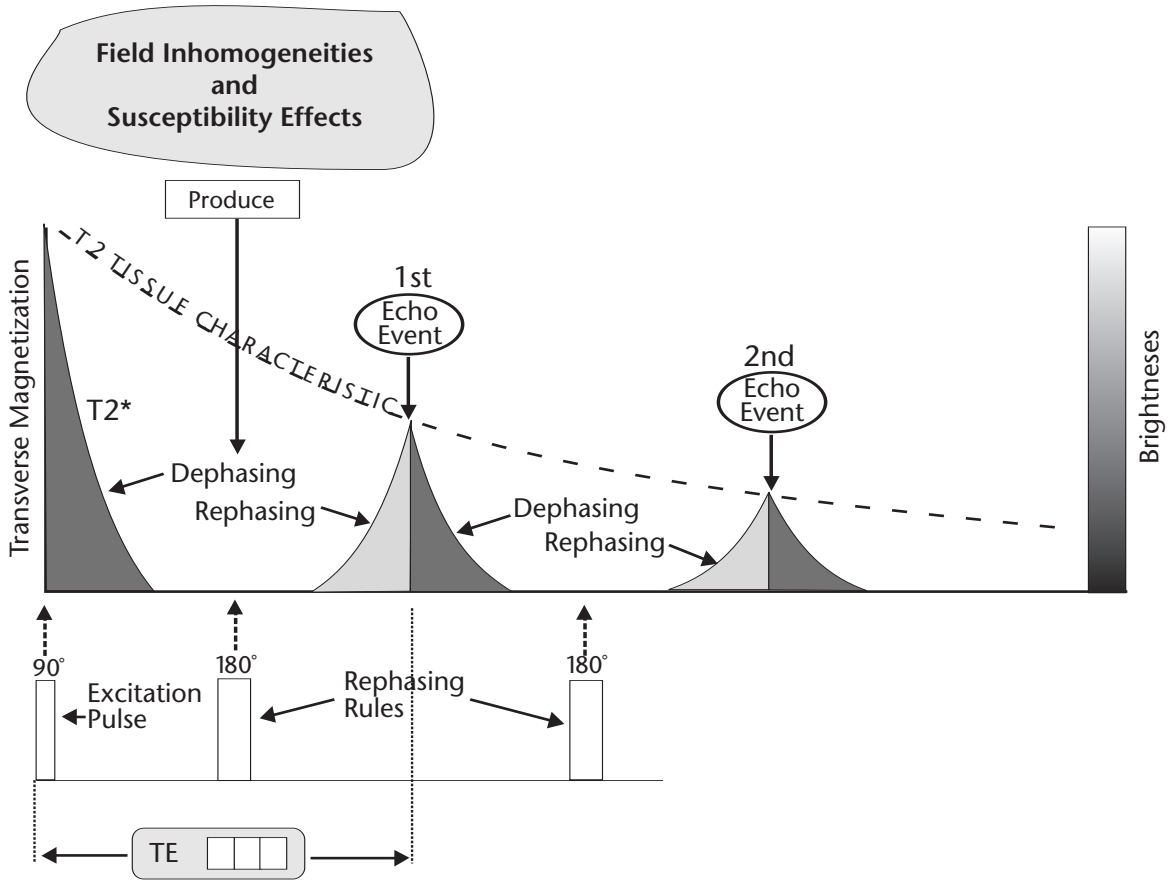


Figure 6-6. Comparison of T1 images produced by spin echo and inversion recovery methods.

develop. A T1 image produced by the inversion recovery method is compared to one produced by the spin echo method in Figure 6-6. Notice the significant difference in contrast. The use of the inversion method for other applications will be discussed in Chapter 8.

Mind Map Summary

Spin Echo Imaging Methods



Spin echo is a technique used to produce an echo event by applying a 180˚ RF pulse to the dephased transverse magnetization. This compensates for the dephasing produced by field inhomogeneities and makes it possible to produce images that show the T2 characteristics of tissue. The time to the echo event, TE, is a protocol factor that can be adjusted to produce different weightings to the T2 contrast. When a short TE value is selected, the T2 effect is reduced, and the resulting image will be either a PD or T1-weighted image, depending on the selected TR value.

It is possible to use a series of 180˚ RF pulses within one cycle to produce multiple echo events, each with a different TE value. Both PD and T2-weighted images can be acquired in the same acquisition.

There is actually a family of spin echo methods that all use the spin echo process to create the echo event. These include the spin echo and inversion recovery methods as well as the GRASE method that uses both spin echoes and gradient echoes in the same acquisition.

7

Gradient Echo Imaging Methods

Introduction And Overview

It is possible to produce an echo event by applying a magnetic field gradient without a 180˚ RF pulse to the tissue as in the spin echo methods. There are several imaging methods that use the gradient echo technique to produce the RF signals and these make up the gradient echo family of methods.

The primary advantage of the gradient echo methods over the spin echo methods is that gradient echo methods perform faster image acquisitions. Gradient echo methods are generally considered to be among the faster imaging methods. They are also used in some of the angiographic applications because gradient echo generally produces bright blood, as we will see in Chapter 12, as well as for functional imaging, as described in Chapter 13. One limitation of the gradient echo methods is they do not produce good T2-weighted images, as will be described later in this chapter. However, by combining the gradient and spin echo methods, this limitation can be overcome.

At this time we will develop the concept of gradient echo and then consider the specific gradient echo imaging methods and their characteristics.

The Gradient Echo Process

Transverse magnetization is present only when a sufficient quantity of protons are spinning in-phase in the transverse plane. As we have

seen, the decay (relaxation) of transverse magnetization is the result of proton dephasing. We also recall that an RF signal is being produced any time there is transverse magnetization and the intensity of the signal is proportional to the level of magnetization.

With the spin echo technique we use an RF pulse to rephase the protons after they have been dephased by inherent magnetic field inhomogeneities and susceptibility effects within the tissue voxel. With the gradient echo technique the protons are first dephased, on purpose, by turning on a gradient and then rephased by reversing the direction of the gradient, as shown in Figure 7-1. A gradient echo can only be created when transverse magnetization is present. This can be either during the free induction decay (FID) period or during a spin echo event. In Figure 7-1 the gradient echo is being created during the FID. Let us now consider the process in more detail.

First, transverse magnetization is produced by the excitation pulse. It immediately begins to decay (the FID process) because of the magnetic field inhomogeneities within each individual voxel. The rate of decay is related to the value of T2*. A short time after the excitation pulse a gradient is applied, which produces a very rapid dephasing of the protons and reduction in the transverse magnetization. This occurs because a gradient is a forced inhomogeneity in the magnetic field. The next step is to reverse the direction of the applied gradient. Even though this is still an inhomogeneity in

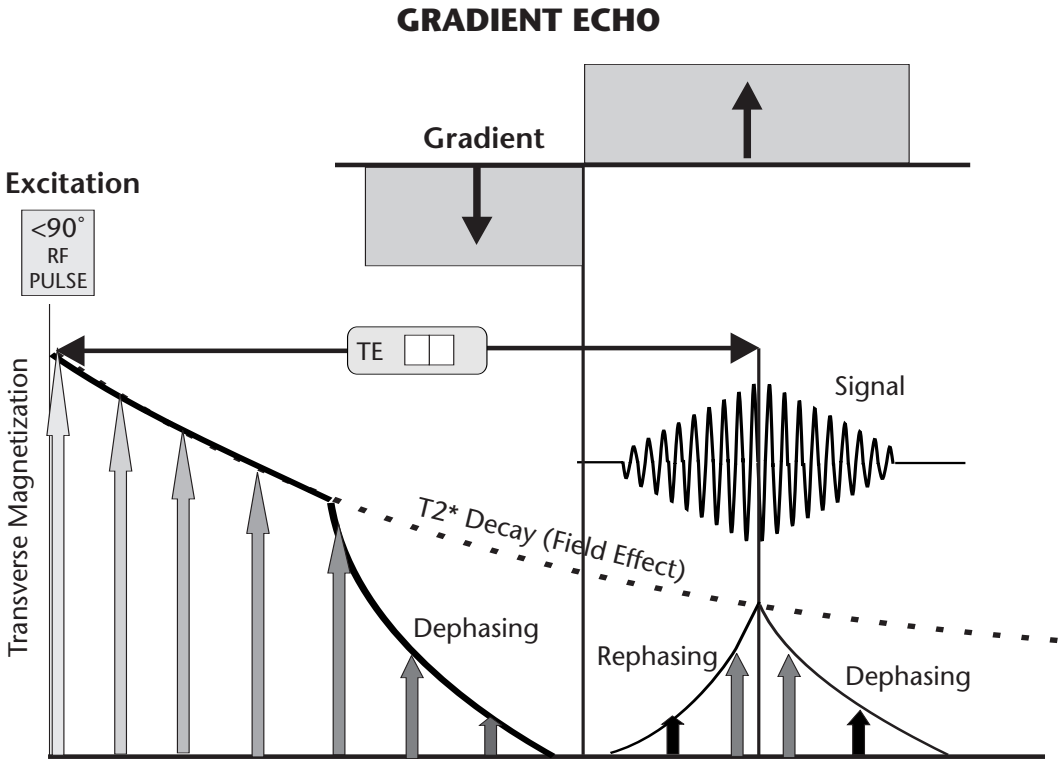


Figure 7-1. The gradient echo process using a magnetic field gradient to produce an echo event during the FID.

the magnetic field, it is in the opposite direction. This then causes the protons to rephase and produce an echo event. As the protons rephase, the transverse magnetization will reappear and rise to a value determined by the FID process. The gradient echo event is a rather well-defined peak in the transverse magnetization and this, in turn, produces a discrete RF signal.

The TE is determined by adjusting the time interval between the excitation pulse and the gradients that produce the echo event. TE values for gradient echo are typically much shorter than for spin echo, especially when the gradient echo is produced during the FID.

Small Angle Gradient Echo Methods

The gradient echo technique is generally used in combination with an RF excitation pulse that has a small flip angle of less than 90˚. We will discover that the advantage of this is that it permits the use of shorter TR values and this, in turn, produces faster image acquisition.

One source of confusion is that each manufacturer of MRI equipment has given his gradient echo imaging methods different trade names. In this text we will use the generic name of *small angle gradient echo* (SAGE) method.

The SAGE method generally requires a shorter acquisition time than the spin echo methods. It is also a more complex method with respect to adjusting contrast sensitivity because the flip angle of the excitation pulse becomes one of the adjustable protocol imaging factors.

Excitation/Saturation-Pulse Flip Angle

We recall that the purpose of the excitation/saturation pulse applied at the beginning of an imaging cycle is to convert or flip longitudinal magnetization into transverse magnetization. When a 90˚ pulse is used, all of the existing longitudinal magnetization is converted into transverse magnetization, as we have seen with the spin echo methods. The 90˚ pulse reduces the longitudinal magnetization to zero (i.e., complete saturation) at the beginning of each imaging cycle. This then means that a relatively long TR interval must be used to allow the longitudinal magnetization to recover to a useful value. The time required for the longitudinal magnetization to relax or to recover is one of the major factors in determining acquisition time. The effect of reducing TR when 90˚ pulses are used is shown in Figure 7-2. As the TR value is decreased, the longitudinal magnetization grows to a lower value and the amount of transverse magnetization and RF signal intensity produced by each pulse is decreased. The reduced signal intensity results in an increase in image noise as described in Chapter 10. Also, the use of short TR intervals with a 90˚ pulse (as in spin echo) cannot produce good PD or T2-weighted images.

One approach to reducing TR and increasing acquisition speed without incurring the disadvantages that have just been described is to use a pulse that has a flip angle of less than 90˚. A small flip-angle (<90˚) pulse converts only a

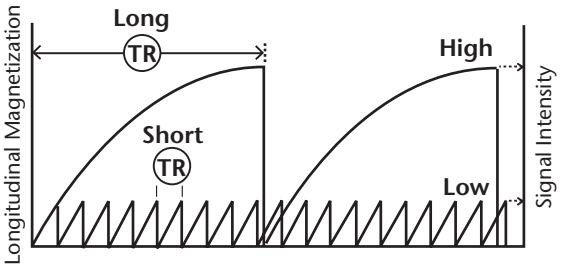


Figure 7-2. The effect of reducing TR on the recovery of longitudinal magnetization within a cycle and the resulting signal intensity when using 90˚ pulses.

fraction of the longitudinal magnetization into transverse magnetization. This means that the longitudinal magnetization is not completely destroyed or reduced to zero (saturated) by the pulse, as shown in Figure 7-3.

Reducing the flip angle has two effects that must be considered together. The effect that we have just observed is that the longitudinal magnetization is not completely destroyed and remains at a relatively high level from cycle to cycle, even for short TR intervals. This will increase RF signal intensity compared to the use of 90˚ pulses. However, as the flip angle is reduced, a smaller fraction of the longitudinal magnetization is converted into transverse magnetization. This has the effect of reducing signal intensity. The result is a combination of these two effects. This is illustrated in Figure 7-4. Here we see that as the flip angle is increased over the range from 0–90˚, the *level* of longitudinal magnetization at the beginning of a cycle decreases. On the other hand, as the angle is increased, the *fraction* of this longitudinal magnetization that is converted into transverse magnetization increases and RF signal intensity increases. The combination of these two

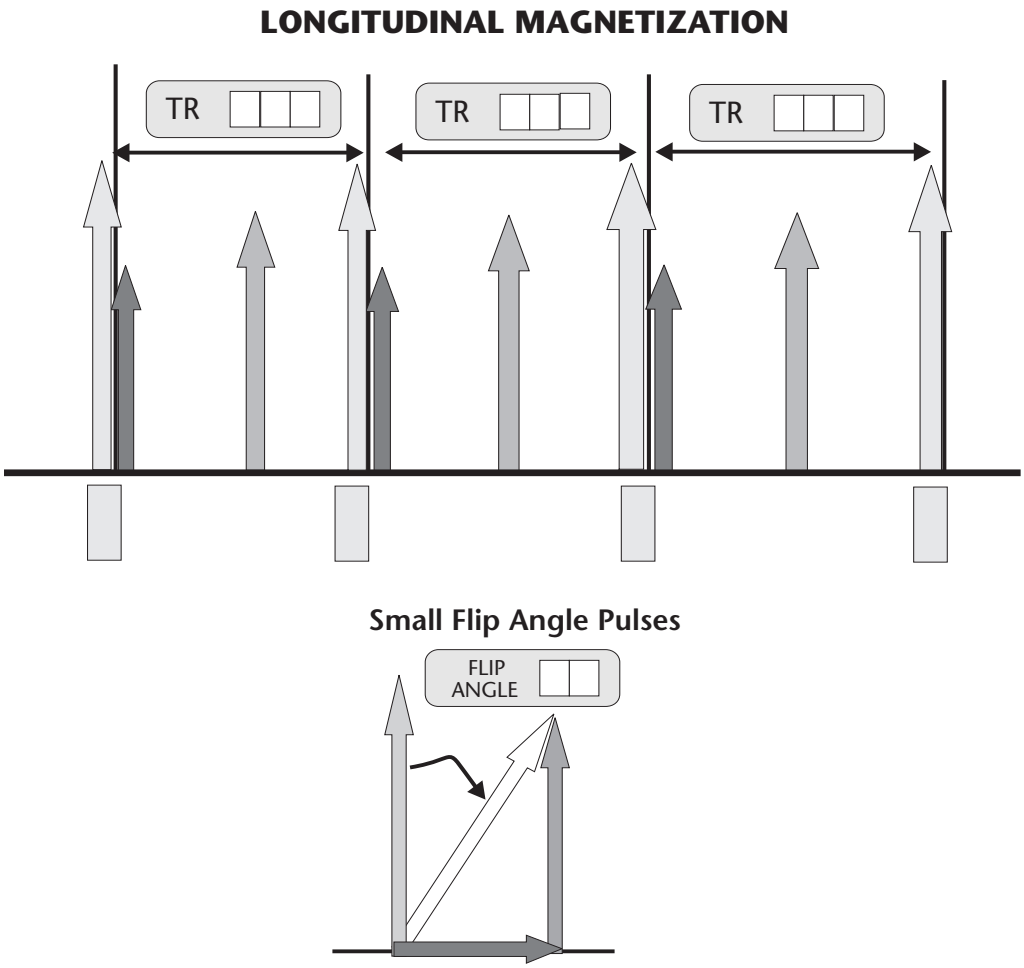


Figure 7-3. The effect of using small flip-angle pulses on longitudinal magnetization.

effects is shown in Figure 7-5. Here we see how changing flip angle affects signal intensity. The exact shape of this curve depends on the specific T1 value of the tissue and the TR interval. For each T1/TR combination there is a different curve and a specific flip angle that produces maximum signal intensity.

Let us now use Figure 7-6 to compare the magnetization of two tissues with different T1 values as we change flip angle. Contrast between the two tissues is represented by the difference in magnetization levels. At this point we are assuming a short TE and considering the contrast associated with only the longitudinal magnetization. The flip-angle range is divided into several specific segments as shown.

Contrast Sensitivity

With the SAGE method the contrast sensitivity for a specific tissue characteristic is controlled by three protocol factors. As with spin echo, TR and TE have an effect. However, the flip angle becomes the factor with the greatest effect on contrast. We will now see how changing flip angle can be used to select specific types of contrast with a basic gradient echo method.

T1 Contrast

Relatively large flip angles (45˚–90˚) produce T1 contrast. This is what we would expect because large flip angles (close to 90˚) and short TR and TE values are similar to the factors used to

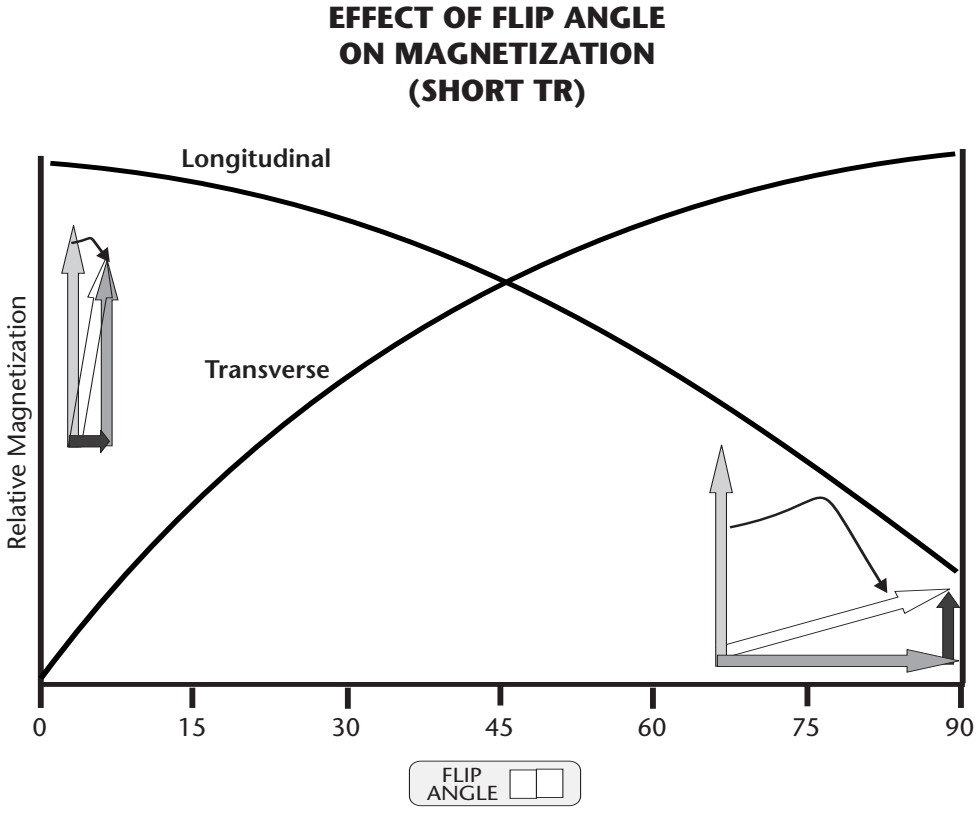


Figure 7-4. The effect of pulse flip angle on the level of both longitudinal and transverse magnetization after the pulse is applied.

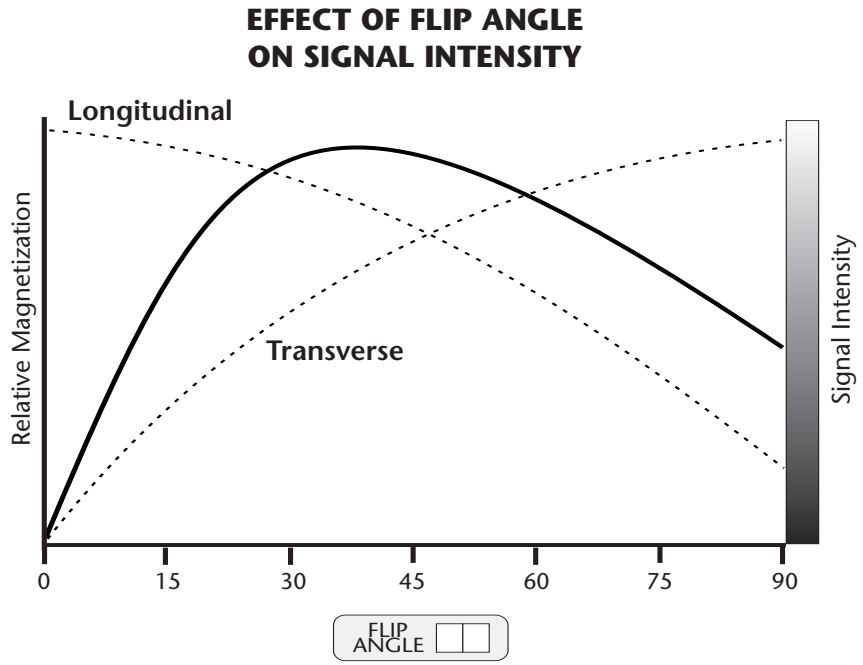


Figure 7-5. The relationship of signal intensity to flip angle.

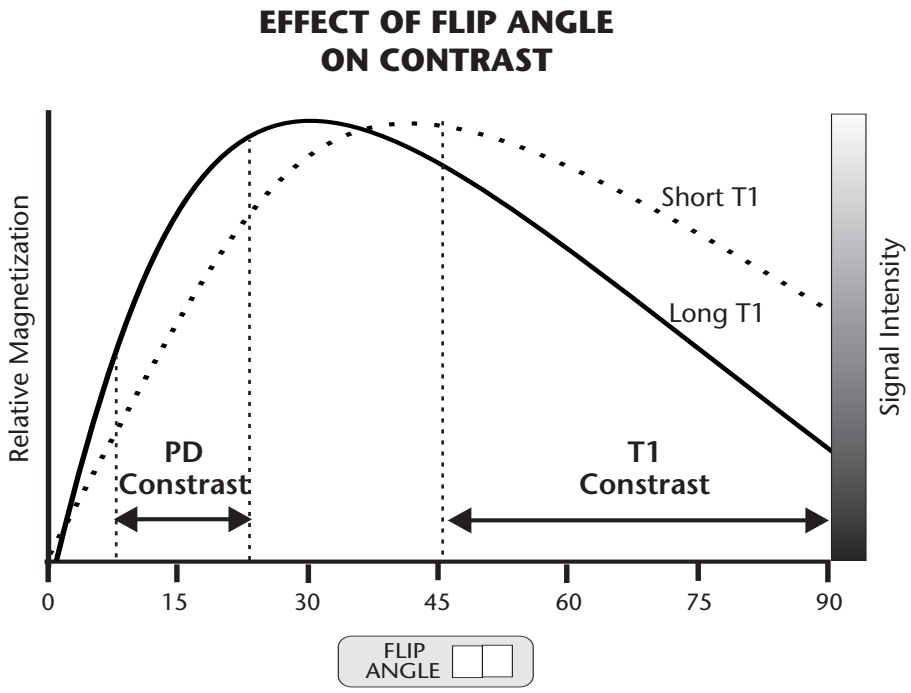


Figure 7-6. The effect of flip angle on contrast.

produce T1 contrast with the spin echo method. Here, with gradient echo, we observe a loss of T1 contrast as the flip angle is decreased significantly from 90˚.

Low Contrast

There is an intermediate range of flip-angle values that produces very little, if any, contrast. This is the region in which the PD and T1 contrast cancel each other for many tissues, such as gray and white matter.

Proton Density (PD) Contrast

Relatively low flip-angle values produce PD contrast. As the flip angle is reduced within this region, there is a significant decrease in magnetization and the resulting signal intensity.

Up to this point we have observed generally how changing the flip angle of the excitation pulse affects signal intensity and contrast. In the SAGE imaging method the flip angle is one of the imaging factors that must be adjusted by the user. However, it becomes somewhat complex because the specific effect of flip angle is modified by the other imaging factors and techniques used to enhance a specific type of contrast.

T2 and T2* Contrast

We recall that T2 contrast is produced by the decay characteristics of transverse magnetization and that there are two different decay rates, T2 and T2*. The slower decay rate is determined by the T2 characteristics of the tissue. The faster decay is produced by small inhomogeneities within the magnetic field often related to variations in tissue susceptibility differences. This decay rate is determined by the T2* of the tissue environment. When a spin echo technique is used, the spinning protons are rephased, and the T2* effect is essentially eliminated. However, when a spin echo technique is not used, the transverse magnetization depends on the T2* characteristics. The gradient echo technique does not compensate for the inhomogeneity and susceptibility effect dephasing as the spin echo technique does. Also, without using a spin echo process the long TE values necessary to produce T2 contrast cannot be achieved. Therefore, a basic gradient echo imaging method is not capable of producing true T2 contrast. The contrast will be determined primarily by the T2* characteristics. The amount of T2* contrast in an image is determined by the selected TE value. In general, longer TE values (but short compared to those used in spin echo) produce more T2* contrast.

Contrast Enhancement

In addition to using combinations of TR, TE, and flip angle to control the contrast characteristics, some gradient echo methods have other features for enhancing certain types of contrast.

When SAGE methods are used with relatively short TR values, there is the possibility that some of the transverse magnetization created in one imaging cycle will carry over into the next cycle. This happens when the TR values are in the same general range as the T2 values of the tissue. SAGE methods differ in how they use the carry-over transverse magnetization.

A typical SAGE sequence is limited to one RF pulse per cycle. If additional pulses were used, as in the spin echo techniques, they would affect the longitudinal magnetization and upset its condition of equilibrium. However, because of the relatively short TR values it is possible for the repeating small-angle excitation pulses to produce a spin echo effect. This can occur only when the TR interval is not much longer than the T2 value of the tissue.

Associated with each excitation pulse, there are actually two components of the transverse magnetization. There is the FID produced by the immediate pulse and a spin echo component produced by the preceding pulses. The spin echo component is related to the T1 characteristics of the tissue. The FID component is related to the T2* characteristics. The contrast characteristics of the imaging method are determined by how these two components are combined. Different combinations are obtained by altering the location of the gradient echo event relative to the transverse magnetization and by turning the spin echo component on or off as described below.

Mixed Contrast

When both the FID and spin echo components are used, an image with mixed contrast characteristics will be obtained. This method produces a relatively high signal intensity compared to the methods described below.

Spoiling and T1 Contrast Enhancement

An image with increased Tl contrast is obtained by suppressing the spin echo component. This is known as *spoiling*. The spin echo component, which is a carryover of transverse magnetization from previous cycles, can be destroyed or spoiled by either altering the phase relationship of the RF pulses or by applying gradient pulses to dephase the spinning protons.

The basic SAGE method discussed up to this point permits faster (than spin echo) image acquisition because the TR can be set to shorter values. However, the gradient echo process can be used in methods that provide fast acquisition based on an entirely different principle. We will now consider methods that achieve their speed by filling many rows of k space during one acquisition cycle.

In Chapter 5 we saw that in the acquisition phase the signal data is being directed into k space from which the image will be reconstructed. The k space is filled one row at a time. The number of rows in the k space for a specific image depends on the required image detail. The process that directs the signals into a specific row of k space is the spatial encoding function performed by one of the gradients. This will be described in Chapter 9. In conventional spin echo and SAGE imaging only one row of k space is filled with each imaging cycle. This is because there is only one echo signal produced per cycle that can be encoded to go to a specific row of k space. This means that the size of k space determines the minimum number of cycles that an acquisition must have. We are about to see some gradient echo methods that can fill many rows of k space in one imaging cycle. This is achieved by using the gradient echo process to produce many echo events from the transverse magnetization that is present during one cycle.

Echo Planar Imaging (EPI) Method

Echo planar is the fast gradient echo imaging method that is capable of acquiring a complete image in a very short time. However, it requires an MRI system equipped with strong gradients that can be turned on and off very rapidly. All systems do not have this capability. The EPI method consists of rapid, multiple gradient echo acquisitions executed during a single spin echo event. The unique characteristic of this method is that each gradient echo signal receives a different spatial encoding and is directed into a different row of k space. The actual spatial encoding process will be described in Chapter 9. Here we are considering only the general concept of EPI and how it achieves rapid acquisition.

The basic EPI method is illustrated in Figure 7-7. Here we see the actions that occur

within one imaging cycle. The cycle usually begins with a spin echo pulse sequence that produces a spin echo event consisting of a period of transverse magnetization as described in Chapter 6. In conventional spin echo imaging we obtain only one signal (and fill one row of k space) from this period of magnetization. What we are about to do with EPI is to chop this one spin echo event into many shorter gradient echo events. The signals from each gradient echo event will receive different spatial encodings and fill different rows of k space.

It is possible to fill all of the k space and acquire a complete image in one cycle. This is described as *single shot* EPI. This is not always practical because it might place some limitations on the image quality that can be achieved and is also very demanding on the gradients. A more practical approach is to divide the acquisition into multiple shots (cycles) with each filling some fraction of the total k space.

The important factor is the number of gradient echo events created in each cycle. This is an adjustable protocol factor and is generally known as the EPI *speed factor*. This is the factor by which the acquisition time is reduced compared to a conventional method using the same TR value.

Gradient And Spin Echo (GRASE) Method

The GRASE method is, as the name indicates, a combination of gradient and spin echo methods. It provides the fast acquisition capability of gradient echo (EPI) with the superior

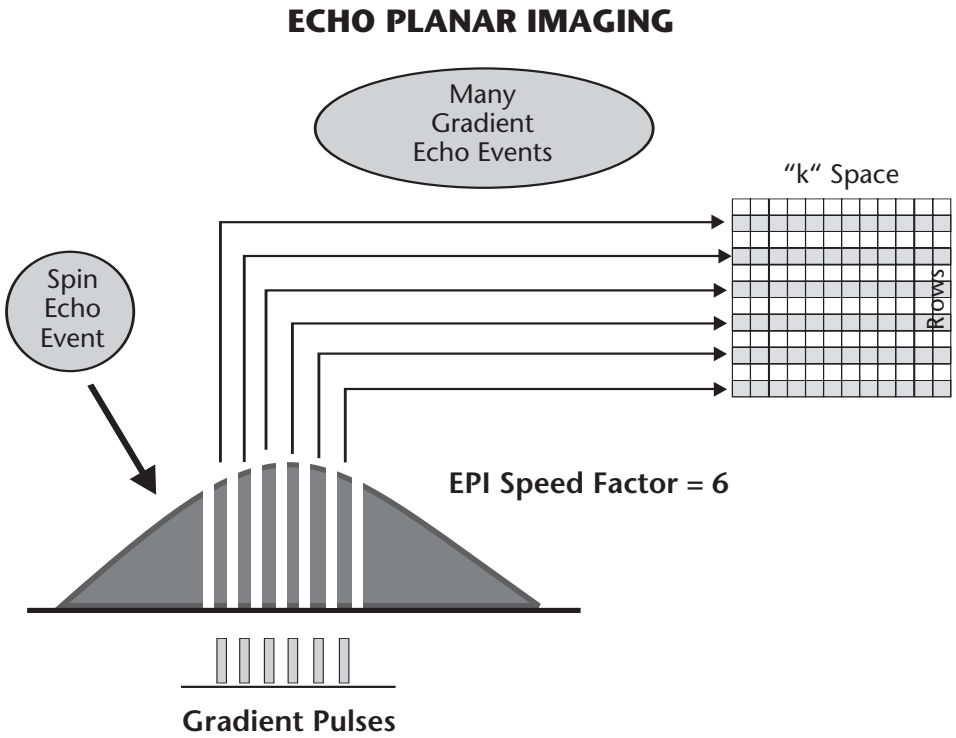


Figure 7-7. The production of many gradient echo events within one imaging cycle with the echo planar imaging (EPI) method.

contrast characteristics of spin echo, including the ability to produce T2 images.

The GRASE method is illustrated in Figure 7-8 where we see the actions occurring within one imaging cycle. The basic cycle is a multiple spin echo as described in Chapter 5. The difference is that in conventional multiple echo, each of the echo events have different TE values and are used to form several images; typically, a PD and a T2 image with the same acquisition. Here the multiple spin echo is used for a different purpose. The multiple spin echoes are used to cover more of k space. As we see, each of the spin echo events is cut into many gradient echoes by the EPI process. This reduces the acquisition time by two factors: the total speed factor is the number of multiple spin echoes multiplied by the EPI speed factor.

Magnetization Preparation

Both SAGE with short TR values and EPI can produce very rapid acquisitions. However, the short time intervals between the gradient echo events do not provide sufficient time for good longitudinal magnetization contrast (T1 or PD) to be formed. This problem is solved by "preparing" the magnetization and forming the contrast just one time at the beginning of the acquisition cycle, as shown in Figure 7-9. Two options are shown.

The longitudinal magnetization is prepared by applying either a saturation pulse, as in the

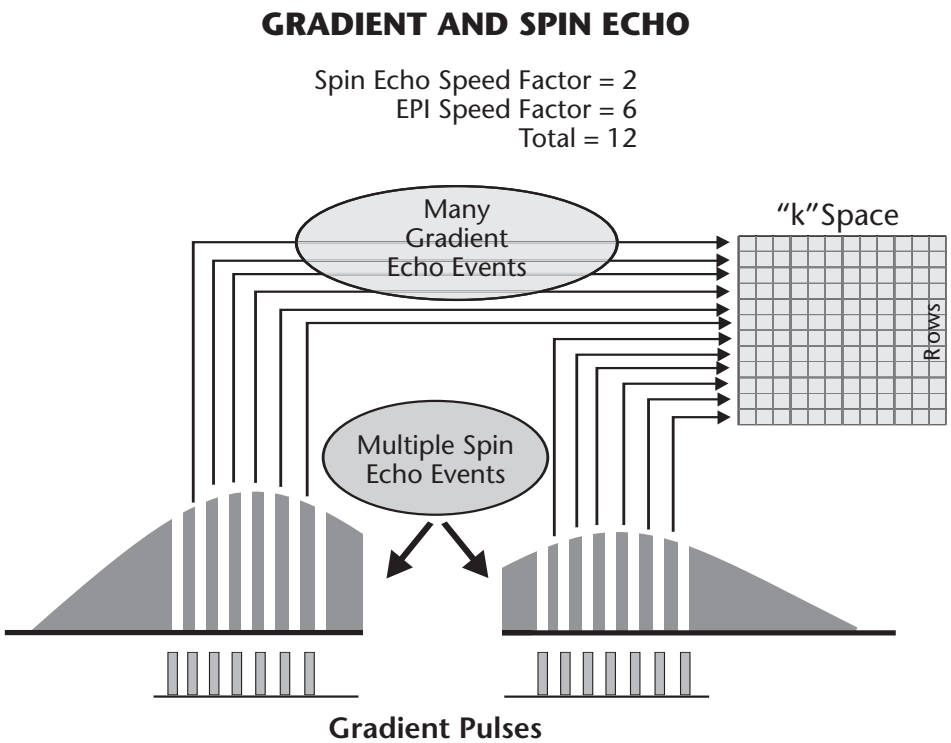


Figure 7-8. The use of the GRASE method to fill many rows of k space and produce a fast acquisition.

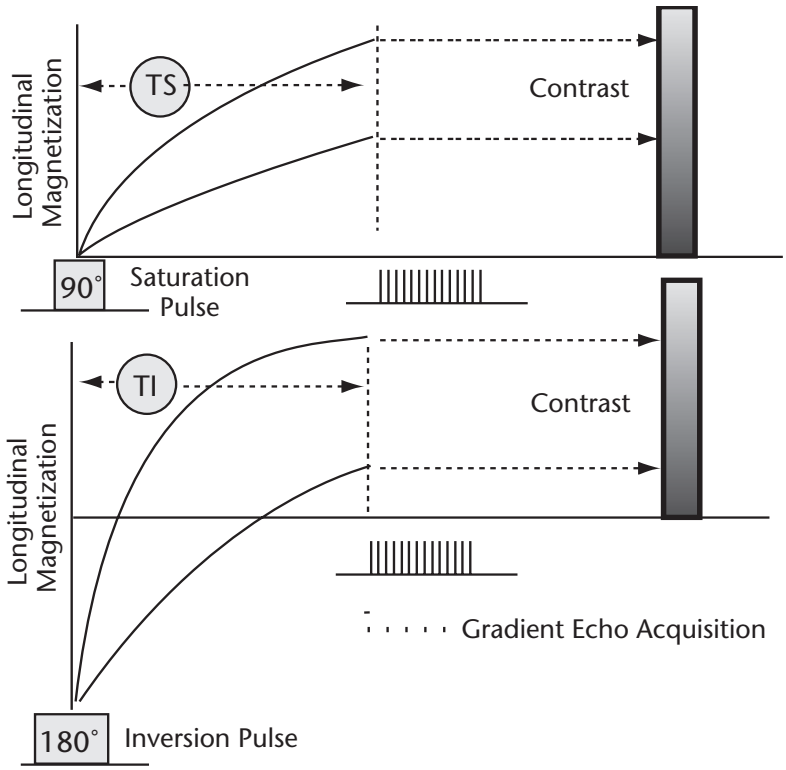
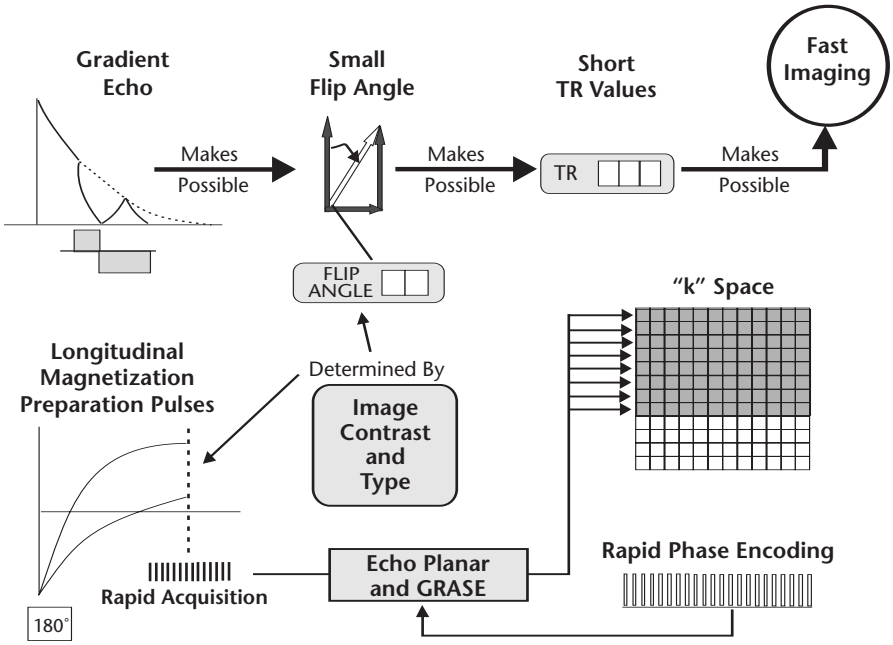


Figure 7-9. Using preparation pulses to produce longitudinal magnetization contrast prior to a rapid gradient echo acquisition.

spin echo method, or an inversion pulse, as in the inversion-recovery method. As the longitudinal magnetization relaxes, contrast is formed between tissues with different T1 and PD values. After a time interval [TI or TS (**T**ime after **S**aturation)] selected by the operator, a rapid gradient echo acquisition begins.

The total acquisition time for this method is the time required by the acquisition cycles plus the TI or TS time interval.

Mind Map Summary
Gradient Echo Imaging Methods



The common characteristic of the gradient echo imaging methods is that a magnetic field gradient is used to produce the echo event rather than a 180˚ RF pulse, as is used in the spin echo methods. One of the principal advantages of the gradient echo process is that it is a relatively fast imaging method.

By using a gradient, and not an RF pulse, to produce the echo event, it is possible to use saturation/excitation pulses with flip angles less than 90˚; thereby all the longitudinal magnetization is not destroyed (saturated) at the beginning of each cycle. Because some longitudinal magnetization carries over from cycle to cycle, it is possible to reduce the TR value and still produce useful signal levels. The reduced TR values result in faster imaging. The flip angle of the RF pulse is an adjustable protocol factor that controls the type of contrast produced.

Echo planar imaging is a gradient echo method in which many echo events, each with a different phase encoding step, are created during each imaging cycle. This makes it possible to fill multiple rows of k space, which results in very fast imaging. GRASE is an imaging method that combines the principles of echo planar and fast (turbo) spin echo to produce rapid imaging acquisitions.

When very fast gradient echo methods are used, there is not sufficient time between the echo events for significant tissue relaxation and contrast to develop. Therefore, the desired contrast is developed at the beginning of the acquisition by applying either inversion or saturation "magnetization preparation" pulses. Then, when the desired contrast has developed, a rapid acquisition is performed.

8

Selective Signal Suppression

Introduction And Overview

There are many times when it is desirable to selectively suppress the signals from specific tissues or anatomical regions. This is done for a variety of reasons including the enhancement of contrast between certain tissues and the reduction of artifacts. During the acquisition process signals can be suppressed based on several properties of a tissue or fluid that make it different from other surrounding tissues. These include differences in T1 values, resonant frequencies, and molecular binding properties. Also, signals from specific anatomical regions can be suppressed or "turned off," usually to prevent interference with imaging in other areas. We will now see how these techniques are used.

Fat and fluid are two materials in the body that can produce very intense signals and brightness in images. This occurs with fat in T1 images and with fluid in T2 images. A possible problem is that these bright regions can reduce the visibility of other tissues and pathologic conditions in the area.

T1-Based Fat And Fluid Suppression

Let us recall that fat has very short T1 values (260 msec) and fluids have very long T1 values (2000 msec). These values are outside of the range of the T1 values of other tissues in the body and are separate and not mixed in with the others. This makes it possible to use T1 as a characteristic for the selective suppression of both fat and fluid.

STIR Fat Suppression

STIR is an inversion recovery method with the TI adjusted to selectively suppress the signals from fat. This uses the fact that fat has a relatively short T1 value and recovers its longitudinal magnetization faster than the other tissues after the inversion pulse. The important point here is that the magnetization of fat passes through the zero level before the other tissues, as shown in Figure 8-1. The TI interval is selected so that the "picture is snapped" by applying the excitation pulse at that time. Because the fat has no magnetization at that time, it will not produce a signal. Since this is achieved with relatively short values for TI, this method of fat suppression is often referred to as **S**hort **T**ime **I**nversion **R**ecovery (STIR).

STIR is just the inversion recovery (IR) method with the TI set to a relatively low value. The description of the basic IR method in Chapter 6 shows how the factor TI is used to select the time at which the longitudinal magnetization "picture is snapped" and the magnetization is converted into image contrast. The ability to use this method to suppress the signals from fat is based on the fact that the longitudinal magnetization of fat passes through zero at a time before and separated from the other tissues. Setting the TI to measure the longitudinal magnetization at the time when fat is at zero produces no signal and fat will be dark in the image.

The best TI value to suppress the signals from fat depends on the T1 value of fat, which

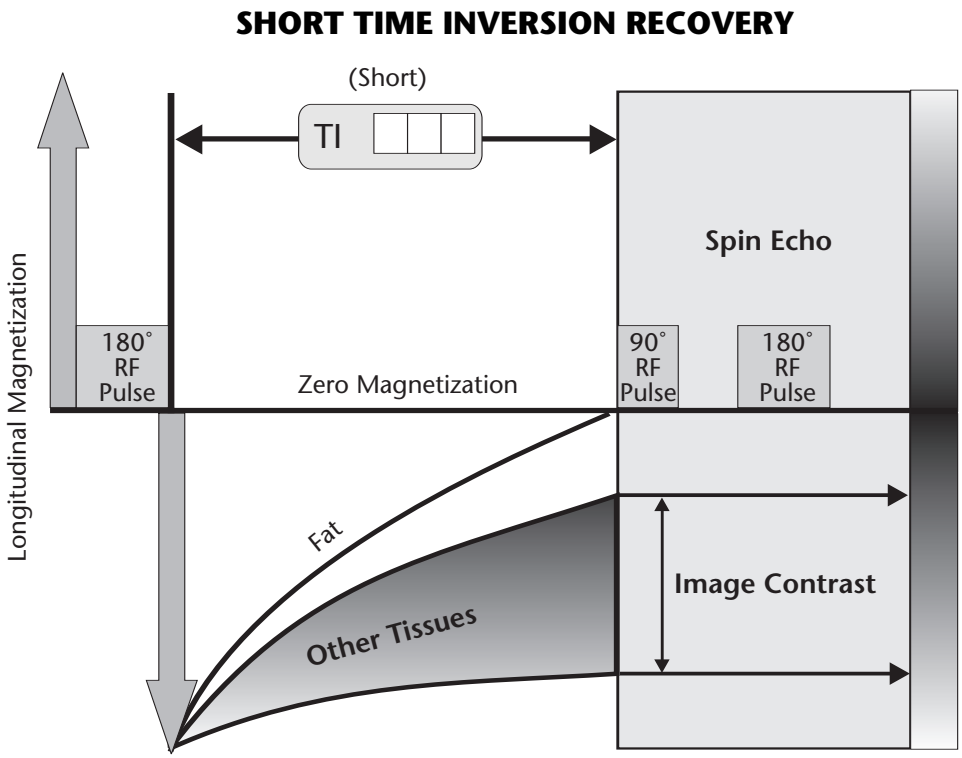


Figure 8-1. The use of STIR to suppress signals from fat by setting TI to a value (short) that will image the longitudinal magnetization at the time when fat is relaxing through the zero level.

depends on the strength of the magnetic field. It will generally be in the range of 120 to 150 msec for field strengths in the 0.5 T to 1.5 T range.

Another consideration with STIR is that the TR must be set relatively long (1500–2000 msec), compared to a T1 image acquisition with spin echo using a TR value of approximately 500 msec. This additional time is required for the longitudinal magnetization to more fully recover after the excitation pulse and before the next cycle can begin.

Fluid Suppression

The suppression of signals from fluids can be achieved by using the IR Method with the TI set to relatively long values as shown in Figure 8-2. This works because the long T1 values of fluids are well separated from the T1 values of other tissues. By setting the TI to a long value as shown, the longitudinal magnetization is converted to transverse and the "picture is snapped" when the fluid is at a zero value. Fluids appear as dark regions in the image. When fluid suppression is used with a T2 image acquisition (long TE), the usually bright fluid is suppressed but other tissues with long T2 values, such as pathologic tissue, remain bright.

Acquisition time is a special concern with this method. That is because when long TI values are used, the TR values must also be long (5000–6000 msec) and that increases the acquisition time. For this reason, the practical

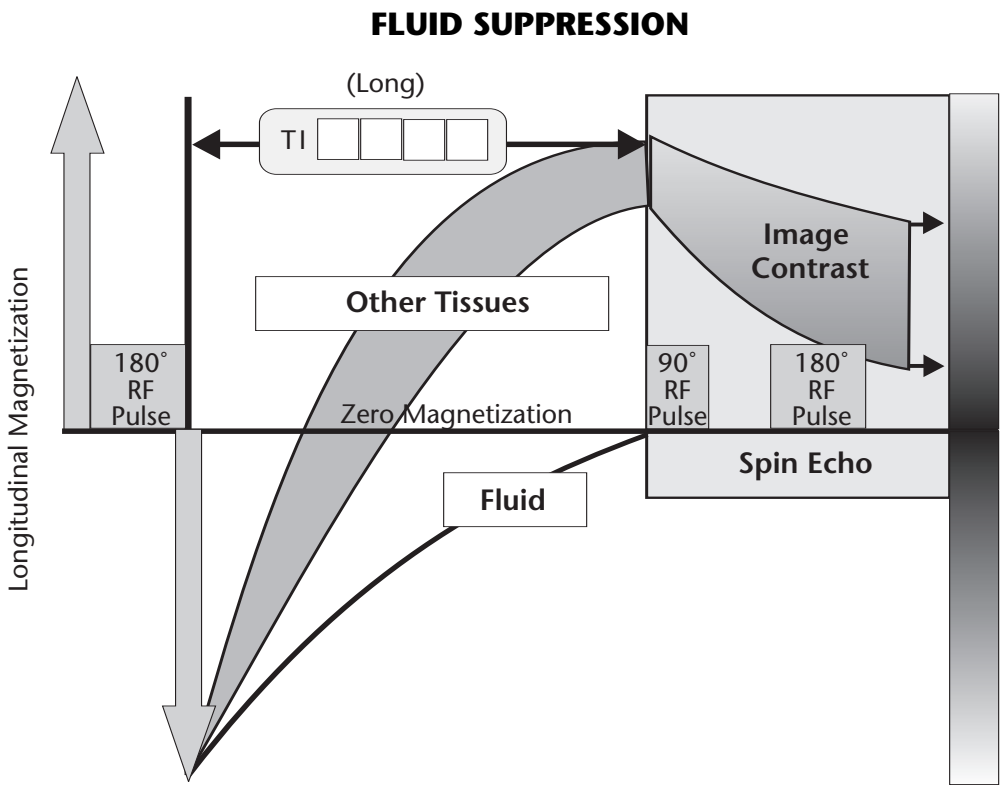


Figure 8-2. The suppression of fluid by selecting a long TI that will image the longitudinal magnetization at the time when fluid is relaxing through zero.

thing is to use this method with one of the fast acquisition techniques.

SPIR Fat Suppression

Spectral **P**resaturation with **I**nversion **R**ecovery (SPIR) is a fat suppression technique which makes use of the fact that fat and the water content of tissues resonate at different frequencies (on the RF frequency spectrum) as described in Chapter 3. We must be careful not to confuse the two fat suppression methods, STIR and SPIR. As we have just seen, STIR uses the difference in *T1 values* to selectively suppress the signals from fat. Now with SPIR, we will use the differences in *resonant frequency* to suppress the fat signals. This technique is illustrated in Figure 8-3. The unique feature of this method is that the imaging cycle begins with an inversion pulse that is applied at the fat resonant frequency. This selectively inverts the longitudinal magnetization of the fat without affecting the other tissues. The TI is set so that the spin echo excitation pulse is applied at the time when the fat longitudinal magnetization is passing through zero. This results in T1 and T2 images with the signals from fat removed.

The advantage of the SPIR method is that the contrast of tissues with relatively short T1 values is not diminished as it might be with the STIR method. For example, the use of gadolinium contrast media reduces the T1 value of the water component of tissue. These

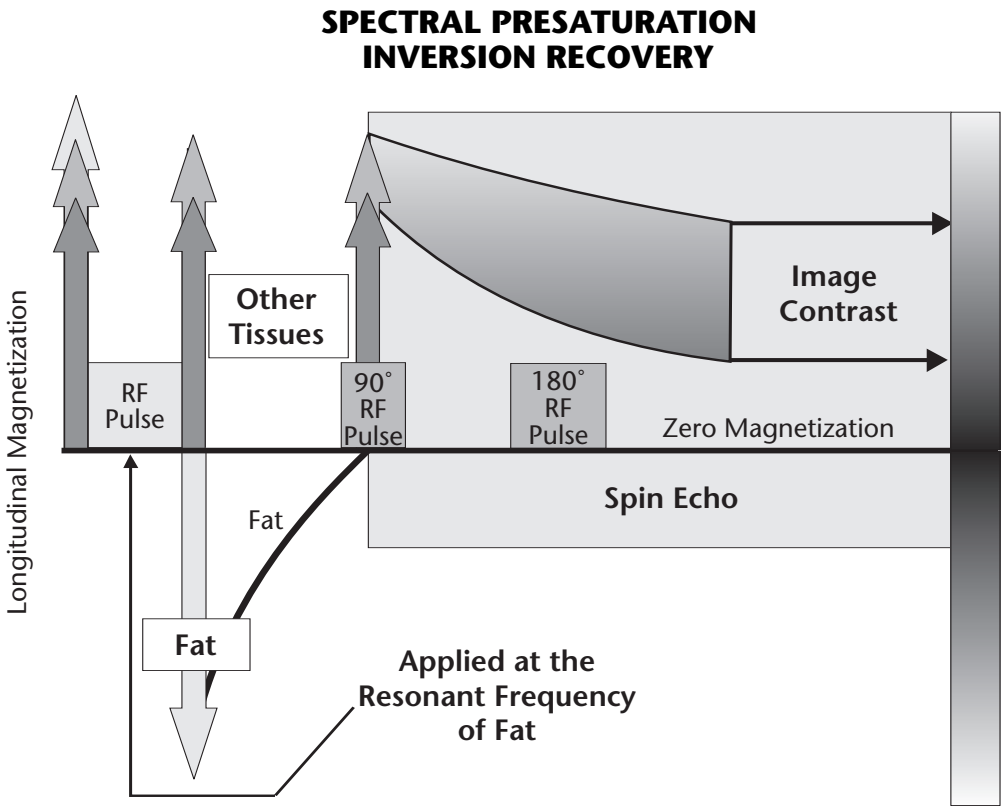


Figure 8-3. Suppressing the signals from fat by applying an inversion pulse tuned to the resonant frequency of fat so that it does not affect the other tissues.

short T1 value signals would be suppressed by STIR, but not by SPIR.

There are some precautions that must be observed when using SPIR. They relate to having very good magnetic field homogeneity. Recall that the resonant frequency is controlled by the field strength in each location. Therefore, for the RF suppression pulse to accurately suppress the fat magnetization over the image area, the fat must be resonating at precisely the same frequency. This requires a very homogeneous (within just a few parts per million) magnetic field. This is achieved by shimming the field before the acquisition, removing metal objects that might distort the field, and by using a relative small field of view.

An alternative to the SPIR method is to apply a saturation rather than an inversion pulse tuned to the fat resonant frequency. This is sometimes referred to as chemical saturation.

Magnetization Transfer Contrast (MTC)

Magnetization Transfer Contrast (MTC) is a technique that enhances image contrast by selectively suppressing the signals from specific tissues. The amount of suppression depends on a specific tissue's *magnetization transfer* characteristics. Maximum suppression is obtained for tissues that have a high level of magnetization transfer.

The MTC technique is illustrated in Figure 8-4. It is based on the principle that the protons in tissue are in different states of mobility, which we will designate as the "free" pool and the "bound" pool.

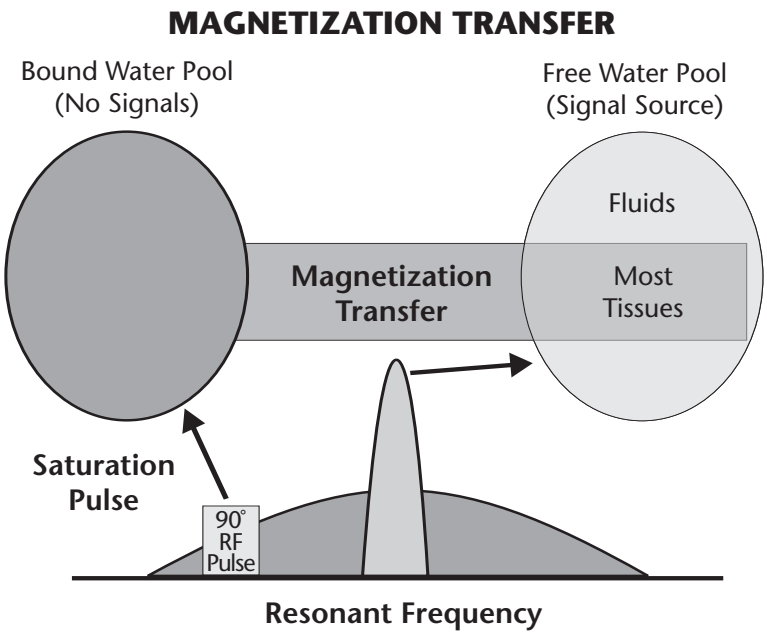


Figure 8-4. The use of magnetization transfer between different types of tissue to suppress selective signals.

Free Proton Pool

The protons that produce signals and are visible in MRI are not rigidly bound and might be considered to be "free" and in a general "semi-solid" structure. This environment produces relatively long T2 values (in comparison to the bound state) and a relatively narrow resonant frequency.

Bound Proton Pool

Most tissues also contain protons that are more rigidly bound and associated with more "solid" structures such as large macromolecules and membranes. These structures have very short T2 values. This means that the transverse magnetization decays before it can be imaged with the usual methods. Therefore, these protons do not contribute to the image. An important characteristic of these protons is that they have a much broader resonant frequency spectrum than the "free" protons.

Magnetization Transfer

Magnetization transfer is a process in which the longitudinal magnetization of one pool influences the longitudinal magnetization in the other pool. In other words, the longitudinal magnetizations of the two pools are coupled together but not to the same degree in all tissues. The MTC process makes use of this difference in coupling to selectively suppress the signals from certain tissues. This is how it is done.

Selective Saturation

The objective of this technique is to saturate and suppress selective signals from specific tissues to increase the contrast.

Prior to the beginning of the imaging acquisition cycle a saturation pulse is applied at a frequency that is different from the resonant frequency of the "free" protons. Therefore, it does not have a direct effect on the protons that are producing the signals. However, the saturation pulse is within the broader resonant frequency of the "bound" protons. It produces saturation of the longitudinal magnetization in the "bound" pool.

The effect of the saturation is now transferred to the longitudinal magnetization of the "free" pool by the magnetization transfer process. The key is that the transfer is not the same for all tissues. Only the tissues with a relatively high magnetization transfer coupling and a significant bound pool concentration will experience the saturation and have their signals reduced in intensity.

Fluids, fat, and bone marrow have very little, if any, magnetization transfer. Therefore, they will not experience the transferred saturation, and will remain relatively bright in the images.

Most other tissues have some, but varying degrees of, magnetization transfer. When the MTC technique is used, the saturation produced by the RF pulse applied to the "bound" protons will be transferred to the "free" protons, but only in those tissues that have a significant magnetization transfer capability. The result is that these tissues will be saturated to some degree and their signal intensities will be reduced.

Therefore, MTC is a way of enhancing contrast in an image by suppressing the signals from tissues that have a relatively high magnetization transfer. One example is to use MTC to reduce the brightness (signal intensity) of brain tissue so that the vascular structures will be brighter in angiography.

Regional Saturation

There are procedures in which it is desirable to suppress signals from specific anatomical regions. The two major applications of this are to reduce motion-induced artifacts, as described in Chapter 14, and to suppress the signals from blood that is flowing in a specific direction, as

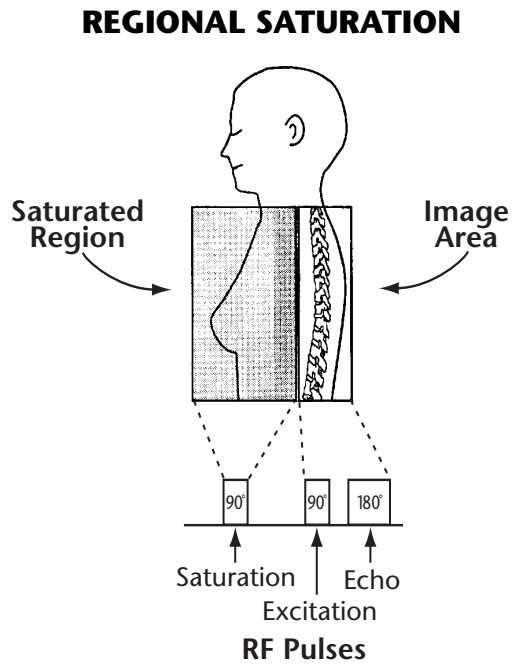


Figure 8-5. The application of a saturation pulse can be directed to a specific anatomical region to suppress undesirable signals from moving tissues.

discussed in Chapter 12. At this time we will consider the general technique, which is illustrated in Figure 8-5.

Let us recall that gradients are used to vary the magnetic field strength across a patient's body. In the presence of a gradient one region of the body is in a different field strength from another and is therefore tuned to a different resonant frequency. This makes it possible to apply RF pulses selectively to specific regions without affecting adjacent regions.

In Chapter 14 we will see that a major source of artifacts in MRI is the motion or movement of tissues and fluids. The motion produces errors in the spatial encoding of the signals that causes them to be displayed in the wrong location in the image. Signals from moving tissues and fluids are displayed as streaks, which are undesirable artifacts.

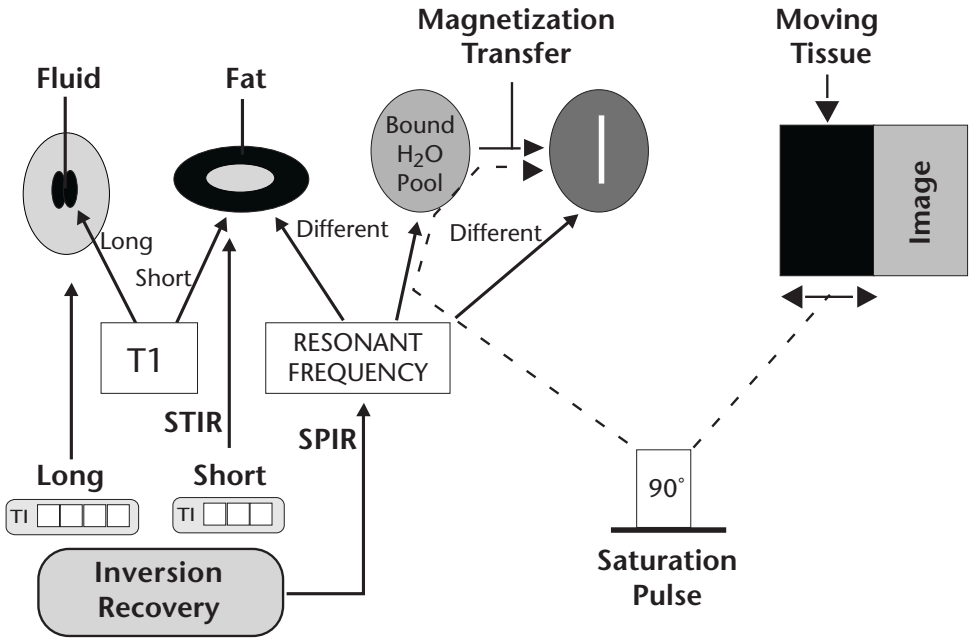
With the regional saturation technique the objective is to suppress selective signals originating from one region, usually the moving tissue or fluid, without affecting these signals in the region that is being imaged. The specific applications of this will be described in Chapter 14.

Prior to the imaging cycle pulse sequence, a saturation pulse is selectively applied to the region that is to be suppressed. The saturation pulse is given a frequency that is different from the frequency of the other imaging pulses. This is so that it will be tuned to the resonant frequency of the region that is to be suppressed. This region will have a resonant frequency different from the imaged area because of the presence of the gradient as described above.

The region that is saturated is a three-dimensional (3-D) volume or slab of tissue. It is important that the slab be properly positioned in relationship to the imaged area for best results.

The application of regional saturation to suppress artifacts will be discussed in more detail when we consider artifacts in Chapter 14.

Mind Map Summary
Selective Signal Suppression



It is often desirable to suppress the signals and resulting brightness of selected tissues or anatomical regions to improve visibility of other tissues or general image quality. It is possible to selectively suppress signals from specific tissues if the tissues are significantly different from the other tissues in terms of some MR characteristic.

Signals from fat, generally very bright in T1 images, can be suppressed with two techniques. Because fat has a very short T1 value compared to other tissues, it can be suppressed with the STIR method, an inversion recovery method in which the TI is set to snap the picture when the magnetization of fat is passing through the zero level. The resonant frequency of fat molecules is slightly different from water molecules because of the chemical shift effect. The SPIR method makes use of this by applying an RF pulse at the fat frequency to reduce the fat magnetization to the zero level at the beginning of each imaging cycle.

Signals from fluid can be suppressed by using an inversion recovery method with the TI set to a long value. This works because fluids have long T1 values and the fluid's magnetization passes through the zero level significantly later and separate from that of tissues. The MTC technique can be used to reduce signal intensity from tissues that have a relatively high magnetization transfer characteristic. This can be used to enhance image contrast.

Saturation pulses can be selectively applied to specific anatomical regions to suppress any signals that could occur from tissues or fluids in that region. This is useful for reducing motion artifacts and also for reducing the signals from flowing blood in specific anatomical regions.

9

Spatial Characteristics of the Magnetic Resonance Image

Introduction And Overview

The MR image formation process subdivides a section of the patient's body into a set of slices and then each slice is cut into rows and columns to form a matrix of individual tissue voxels. This was introduced first in Chapter 1 and illustrated in Figure 1-3. The RF signal from each individual voxel must be separated from all of the other voxels and its intensity displayed in the corresponding image pixel, as shown in Figure 9-1. This is achieved by encoding or addressing the signals during the acquisition phase and then, in effect, delivering the signal intensities to the appropriate pixels which have addresses within the image during the reconstruction phase. Because there are two dimensions, or directions, in an image, two different methods of encoding must be used. This is analogous to mail that must have both a street name and a house number in the address. We are about to see that the two methods of addressing the signals are called *frequency-encoding* and *phase-encoding*. One method is applied to one direction in the image and the other method is used to address in the other direction.

This two-step process consisting of the *signal acquisition* phase followed by the *image reconstruction* phase is illustrated in Figure 9-2. Different actions happen in these two phases that must be considered when setting up an imaging procedure.

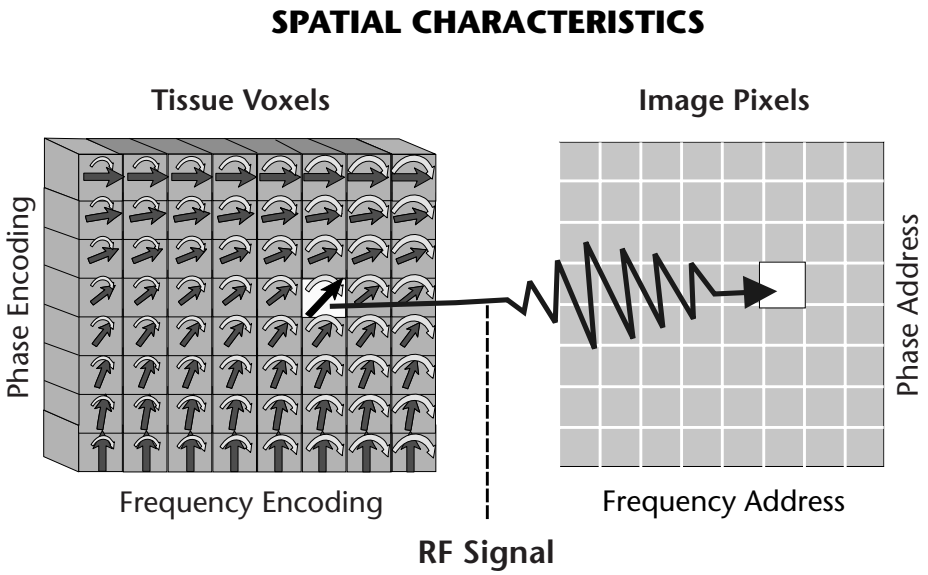


Figure 9-1. The relationship of tissue voxels to image pixels.

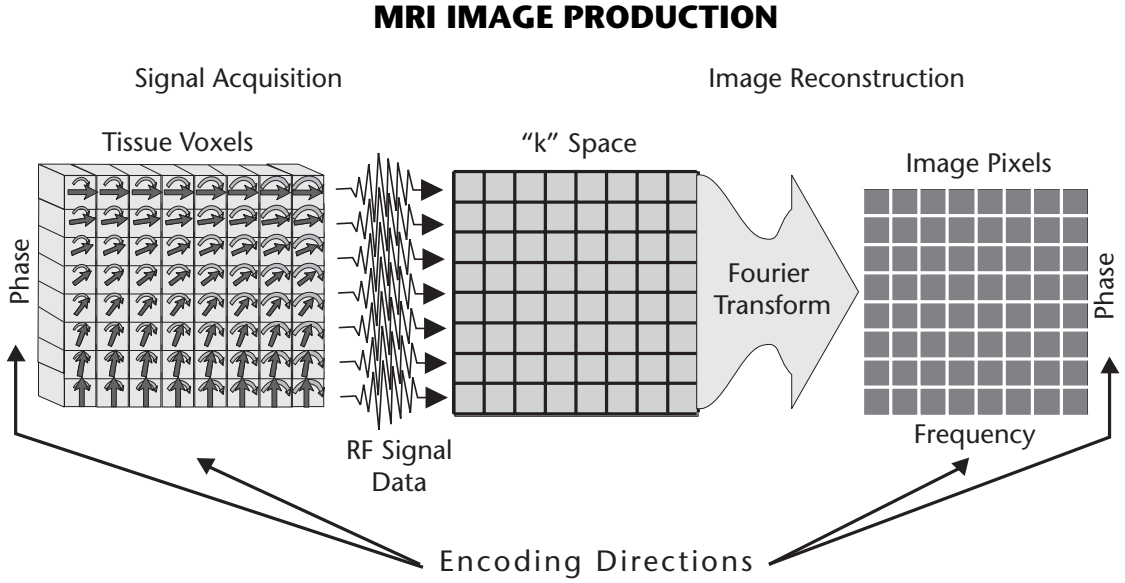


Figure 9-2. The two phases—signal acquisition and image reconstruction—that are required to produce an MR image.

Signal Acquisition

During the acquisition phase the RF signals are emitted by the tissue and received by the RF coils of the equipment. During this process the signals from the different slices and voxels are given distinctive frequency and phase characteristics so that they can be separated from the other signals during image reconstruction. The acquisition phase consists of an imaging cycle that is repeated many times. The time required for image acquisition is determined by the time TR, which is the duration of one cycle or its repetition time, and the number of cycle repetitions. The number of cycles is determined by the image quality requirements. In general, the quality of an image can be improved by increasing the number of acquisition cycles. This is considered in much more detail in Chapter 10.

The result of the image acquisition process is a large amount of data collected and stored in computer memory. At this point the data represent RF signal intensities characterized by the two characteristics, *frequency* and *phase*. The concept of frequency and phase will be developed later. At this point in the process the data are not yet in the form of an image but are located in k space. The data will later be transformed into image space by the reconstruction process.

Image Reconstruction

Image reconstruction is a mathematical process performed by the computer. It transforms the data collected during the acquisition phase into an image. We can think of reconstruction as the process of sorting the signals collected during the acquisition and then delivering them to the appropriate image pixels. The mathematical process used is known as *Fourier transformation*. Image reconstruction is typically much faster than image acquisition and requires very little, if any, control by the user.

Image Characteristics

The most significant spatial characteristic of an image is the size of the individual tissue voxels. Voxel size has a major effect on both the detail and noise characteristics of the image. The user can select the desired voxel size by adjusting a combination of imaging factors, as described in Chapter 10.

Gradients

The spatial characteristics of an MR image are produced by actions of the gradients applied during the acquisition phase. Magnetic field gradients are used first to select slices and then give the RF signals the frequency and phase characteristics that create the individual voxels.

As we will see later, a gradient in one direction is used to create the slices, and then gradients in the other directions are used to cut the slices into rows and columns to create the individual voxels. However, these functions can be interchanged or shared among the different gradient coils to permit imaging in any plane through the patient's body.

The functions performed by the various gradients usually occur in a specific sequence. During each individual image acquisition cycle the various gradients will be turned on and off at specific times. As we will see later, the gradients are synchronized with other events such as the application of the RF pulses and the acquisition of the RF signals.

Slice Selection

There are two distinct methods used to create the individual slices. The method of *selective excitation* actually creates the slice during the acquisition phase. An alternative method is to acquire signals from a large volume of tissue (like an organ) and then create the slices during

the reconstruction process. These are often referred to as 2-D (volume) and 3-D (volume) acquisitions. However, each produces data that are reconstructed into slice images. Both methods have advantages and disadvantages, which will be described later.

Selective Excitation

The first gradient action in a cycle defines the location and thickness of the tissue slice to be imaged. We will illustrate the procedure for a conventional transaxial slice orientation. Other orientations, such as sagittal, coronal, and angled combinations, are created by interchanging and combining gradient directions.

Slice selection using the principle of selective excitation is illustrated in Figure 9-3. When a magnetic field gradient is oriented along the patient axis, each slice of tissue is in a different field strength and is tuned to a different resonant frequency. Remember, this is because the resonant frequency of protons is directly proportional to the strength of the magnetic field at the point where they are located. This slice selection gradient is present whenever RF pulses are applied to the body. Since RF pulses contain frequencies within a limited range (or bandwidth), they can excite tissue only in a specific slice. The location of the slice can be changed or moved along the gradient by using a slightly different RF pulse frequency. The thickness of the slice is determined by a combination of two factors: (1) the strength, or steepness, of the gradient, and (2) the range of frequencies, or bandwidth, in the RF pulse.

Multi-Slice Imaging

In most clinical applications, it is desirable to have a series of images (slices) covering a specific anatomical region. By using the multi-slice mode, an entire set of images can be acquired simultaneously. The basic principle is illustrated in Figure 9-4. The slices are separated by applying the RF pulses and detecting the signals from the different slices at different times, in sequence, during each imaging cycle.

When the slice selection gradient is turned on, each slice is tuned to a different resonant frequency. A specific slice can be selected for

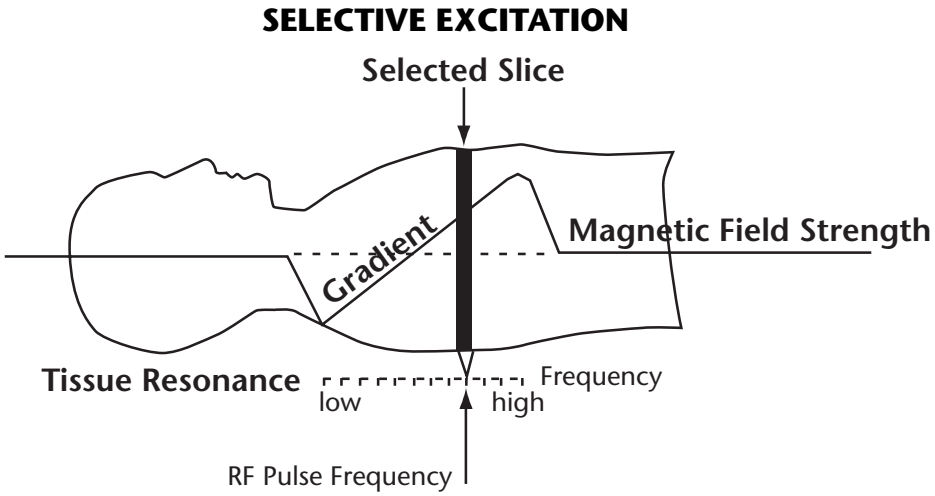


Figure 9-3. The use of a gradient to tune a specific slice so that it can be selectively excited by an RF pulse.

excitation by adjusting the RF pulse frequency to correspond to the resonant frequency of that slice. The process begins by applying an excitation pulse to one slice and collecting the echo signal. Then, while that slice undergoes longitudinal relaxation before the next cycle can begin, the excitation pulse frequency is shifted to excite another slice. This process is repeated to excite and collect signals from the entire set of slices at slightly different times within one TR interval.

The advantage of multi-slice imaging is that a set of slices can be imaged in the same time as a single slice. The principal factor that limits the number of slices is the value of TR. It takes a certain amount of time to excite and then collect the signals from each slice. The maximum number of slices is the TR value divided by the time required for each slice. This limitation is especially significant for T1-weighted images that use relatively short TR values.

A factor to consider when selecting the slicing mode is that multiple slice selective excitation cannot produce the contiguous slices that the volume acquisition technique can. With selective excitation there is the possibility that when an RF excitation/saturation pulse is applied to one slice of tissue, it will also produce some effect in an adjacent slice. This is a reason for leaving gaps between slices during the acquisition.

Volume Acquisition

Volume (3-D) image acquisition has the advantage of being able to produce thinner and more contiguous slices. This is because of the process used to slice the tissue. Rather than producing each slice during the acquisition phase, the slicing is done during the reconstruction phase using the process of phase-encoding. The actual process of phase-encoding will be described later in this chapter. At this time we only consider how it is used for slicing. With this method, no gradient is present when the RF pulse is applied to the tissue. Since all tissue within an anatomical region, such as the head, is tuned to the same resonant frequency, all tissues are

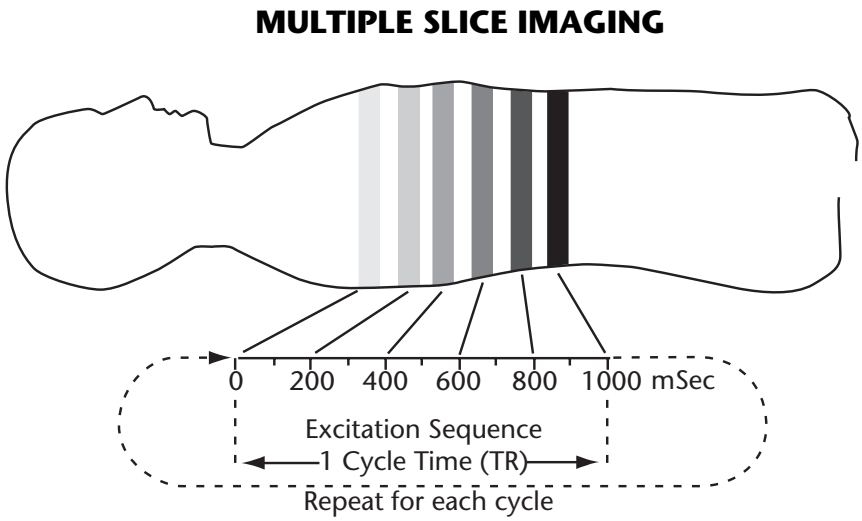


Figure 9-4. Multiple slice imaging applies pulses to and produces signals from different slices within one imaging cycle.

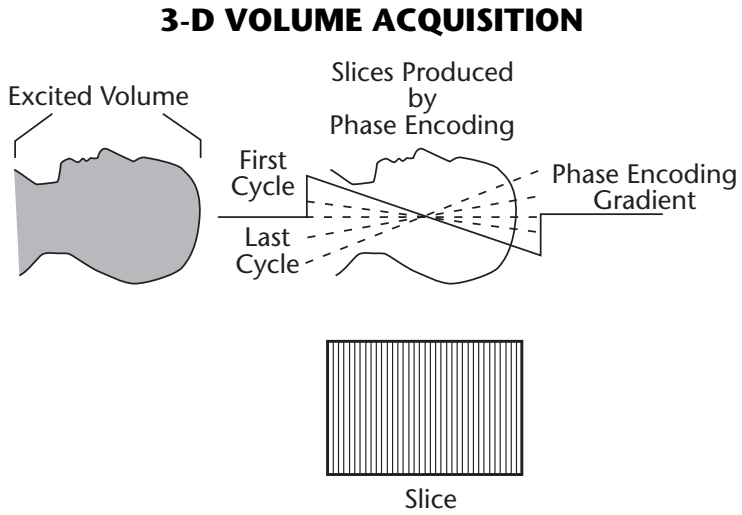


Figure 9-5. The 3-D volume acquisition process uses the phase-encoding process to produce thin slices.

excited simultaneously. The next step, as illustrated in Figure 9-5, is to apply a phase-encoding gradient in the slice selection direction. In volume imaging, phase-encoding is used to create the slices in addition to creating the voxel rows as described below. The phase-encoding gradient used to define the slices must be stepped through different values, corresponding to the number of slices to be created. At each gradient setting, a complete set of imaging cycles must be executed. Therefore, the total number of cycles required in one acquisition is multiplied by the number of slices to be produced. This has the disadvantage of causing 3-D volume acquisitions to have a relatively long acquisition time compared to 2-D multiple slice acquisitions. That is why this type of acquisition is often used with one of the faster imaging methods.

The primary advantage of volume imaging is that the phase-encoding process can generally produce thinner and more contiguous slices than the selective excitation process used in 2-D slice acquisition. The primary disadvantage is longer acquisition times.

Frequency Encoding

A fundamental characteristic of an RF signal is its frequency. Frequency is the number of cycles per second of the oscillating signal. The frequency unit of Hertz (Hz) corresponds to one cycle per second. Radio broadcast stations transmit signals on their assigned frequency. By tuning our radio receiver to a specific frequency we can select and separate from all other signals the specific broadcast we want to receive. In other words, the radio broadcasts from all of the stations in a city are frequency encoded. The same process (frequency-encoding) is used to cause voxels to produce signals that are different and can be used to create one dimension of the image.

Let us review the concept of RF signal production by voxels of tissue, as shown in Figure 9-6. RF signals are produced only when transverse magnetization is present. The unique characteristic of transverse magnetization that produces the signal is a spinning magnetic effect, as shown. The transverse magnetization spins around the axis of the magnetic

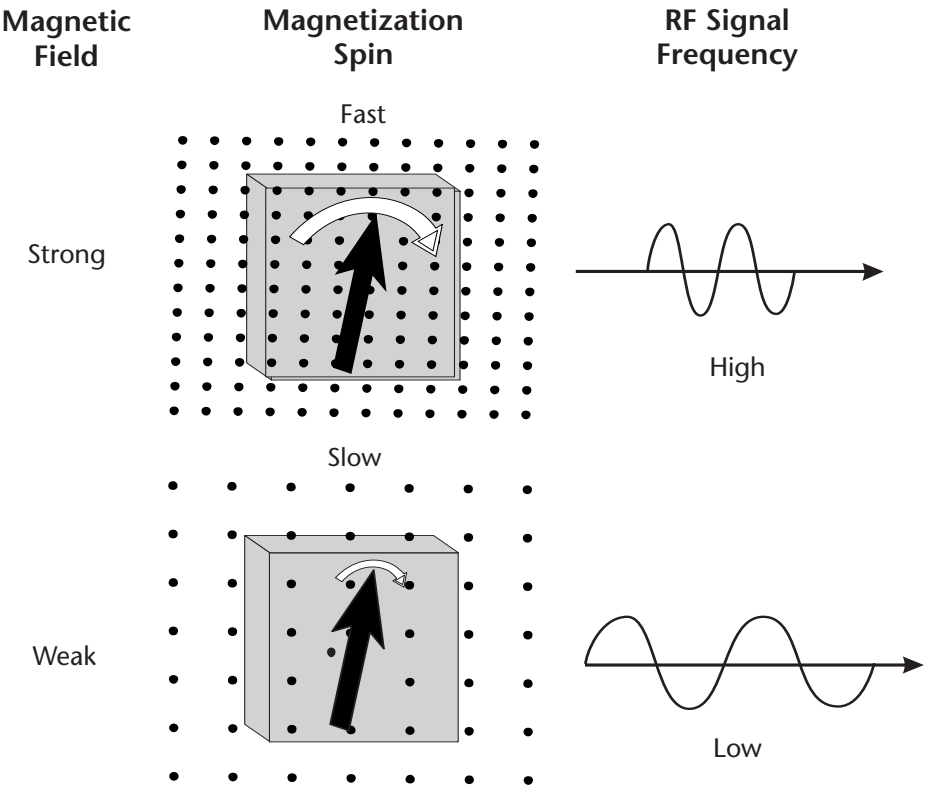


Figure 9-6. The effect of field strength on the frequency of RF signals produced by transverse magnetization.

field. A spinning magnet or magnetization in the vicinity of a coil forms a very simple electric generator. It generates one cycle for each revolution of the magnetization. When the magnetization is spinning at the rate of millions of revolutions per second, the result is an RF signal with a frequency in the range of Megahertz (MHz).

Resonant Frequency

The frequency of the RF signal is determined by the spinning rate of the transverse magnetization. This, in turn, is determined by two factors, as was described in Chapter 3. One factor is the specific magnetic nuclei (usually protons) and the other is the strength of the magnetic field in which the voxel is located. When imaging protons, the strength of the magnetic field is the factor used to vary the resonant frequency and the corresponding frequency of the RF signals. In Figure 9-6 we see two voxels located in different strength fields. The result is that they produce different frequency signals.

Figure 9-7 shows the process of frequency encoding the signals for a row of voxels. In this example, a gradient is applied along the row. The magnetic field strength is increased from left to right. This means that each voxel is located in a different field strength and is resonating at a frequency different from all of the others. The resonant and RF signal frequencies increase from the left to right as shown.

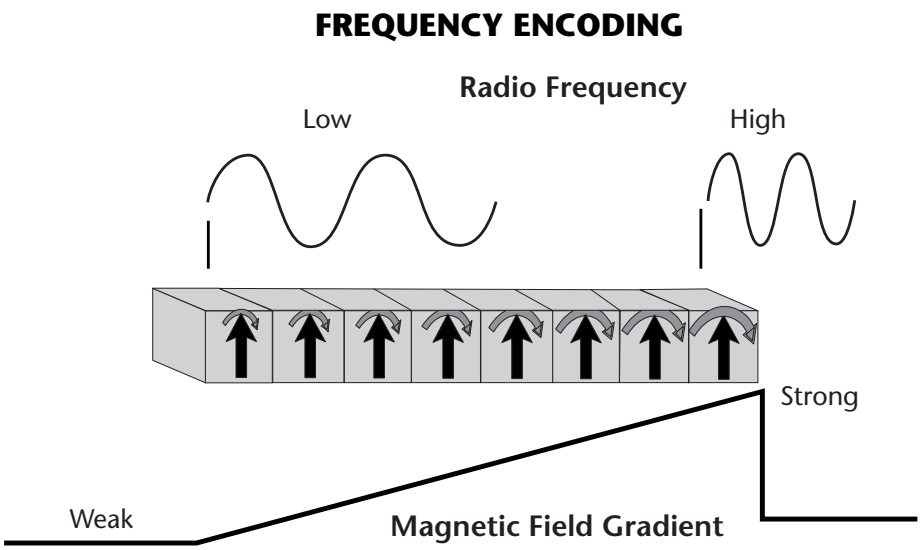


Figure 9-7. The frequency encoding of a row of voxels within a slice.

The frequency-encoding gradient is on at the time of the echo event when the signals are actually being produced. The signals from all of the voxels in a slice are produced simultaneously and are emitted from the body mixed together to form a composite signal at the time of the echo event. The individual signals will be separated later by the reconstruction process to form the voxels.

Phase-Encoding

Phase is a relationship between one signal and another, as illustrated in Figure 9-8. Here we see two voxels producing RF signals. The transverse magnetization is spinning at the same rate and producing signals that have the same frequency. However, we notice that one signal is more advanced in time or is out of step with the other. In other words, the two signals are out of phase. The significance of voxel-to-voxel phase in MRI is that it can be used to separate signals and create one dimension in the image.

A phase difference is created by temporarily changing the spinning rate of the magnetization of one voxel with respect to another. This happens when the two voxels are located in magnetic fields of different strengths. This can be achieved by turning on a gradient, as shown in Figure 9-9.

Let us begin the process of phase encoding by considering the column of voxels shown in the illustration. We are assuming that all voxels have the same amount of transverse magnetization and that the magnetization is spinning in-phase at the time just prior to the phase-encoding process.

When the phase-encoding gradient is turned on, we have the condition illustrated with the center column of voxels. The strength of the magnetic field is increasing from bottom to top. Therefore, the magnetization in each voxel is spinning at a different rate with the speed increasing from bottom to top. This causes the magnetization from voxel to voxel to get out of step or produce a phase difference. The phase-encoding gradient remains on for a short period of time and then is turned off. This leaves the condition represented by the

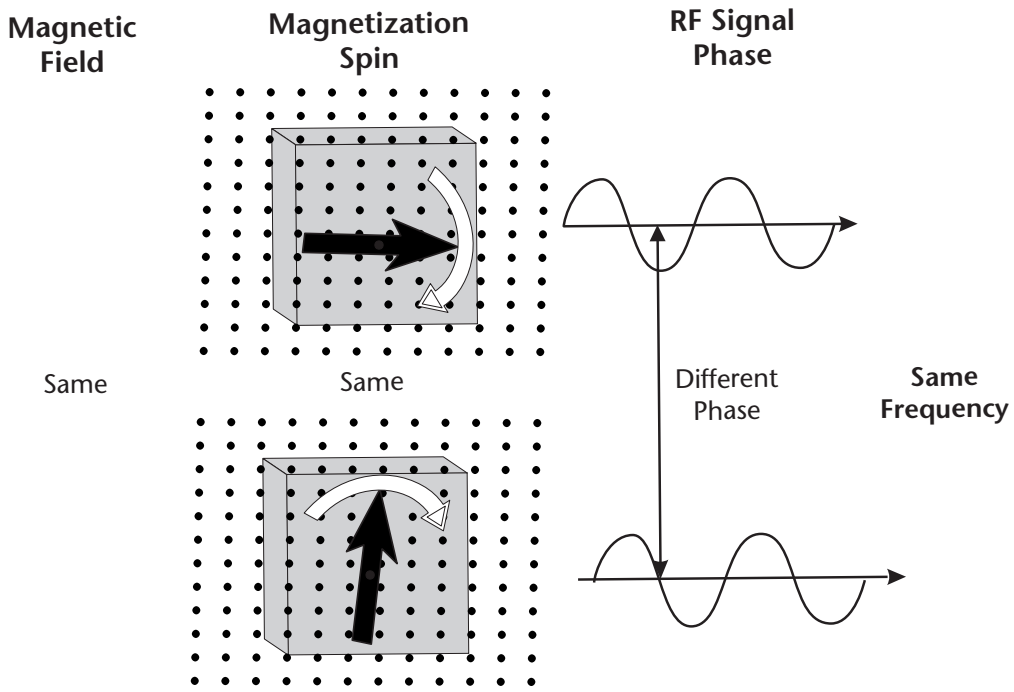


Figure 9-8. The concept of phase between the signals from two voxels.

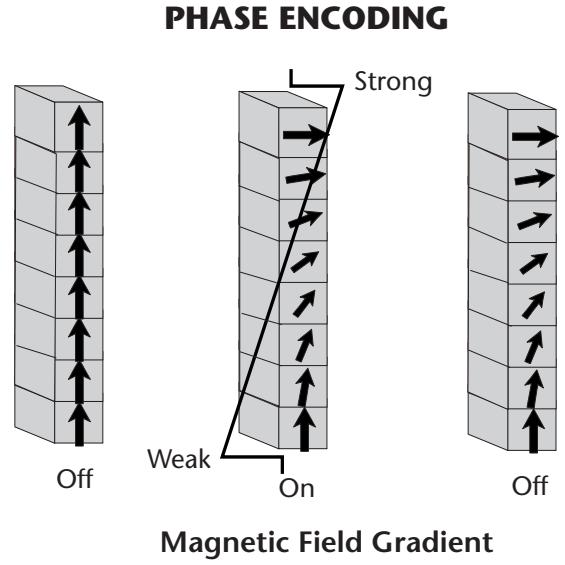


Figure 9-9. Phase-encoding produced by turning on a gradient for a short time, and then turning it off. The phase difference remains.

column of voxels on the right. This is the condition that exists at the time of the echo event when the signals are actually produced. As we see, the signals from the individual voxels are different in terms of their phase relationship. In other words, the signals are phase-encoded. All of the signals are emitted at the same time and mixed together as a composite echo signal. Later, the reconstruction process will sort the individual signal components.

Phase-encoding is the second function performed by a gradient during each cycle, as shown in Figure 9-10. During each pass through an imaging cycle, the phase-encoding gradient is stepped to a slightly different value.

The signals acquired with each phase-encoding *gradient strength* fills one row of k space. This is a very important point that should be emphasized: *Each row of k space is reserved for signals with a specific degree of phase-encoding*. The degree of phase-encoding is determined by the strength and duration of the phasing gradient applied during each cycle. Therefore, the phase-encoding process must be repeated depending on the size of k space and that is determined by the image matrix size in the phase-encoded direction.

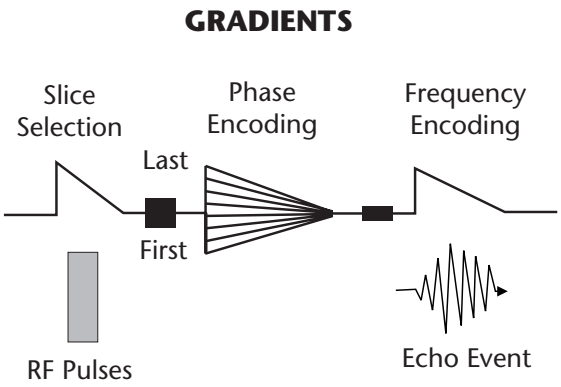


Figure 9-10. The relationship of the three gradient actions—slice selection, phase-encoding, and frequency-encoding—to each other and to the RF pulses and signals. They are applied in different directions.

One MRI phase-encoding step produces a composite signal from all voxels within a slice. The difference from one step to another is that individual voxel signals have a different phase relationship within the composite signal.

To reconstruct an image by the conventional 2-D Fourier transformation method, one composite signal, or phase-encoded step, must be collected for each voxel to be created in the phase-encoding direction. Therefore, the minimum number of steps required to produce an image is determined by the size of the image matrix and k space. It takes 256 phase-encoding steps to produce an image with a 256×256 matrix.

The Gradient Cycle

We have seen that various gradients are turned on and off at specific times within each imaging cycle. The relationship of each gradient to the other events during an imaging cycle is shown in Figure 9-10. The three gradient activities are:

1. The slice selection gradient is on when RF pulses are applied to the tissue. This limits magnetic excitation, inversion, and echo formation to the tissue located within the specific slice.
2. The phase-encoding gradient is turned on for a short period in each cycle to produce a phase difference in one dimension of the image. The strength of this gradient is changed slightly from one cycle to another to fill the different rows of k space needed to form the image.
3. The frequency-encoding gradient is turned on during the echo event when the signals are actually emitted by the tissue. This causes the different voxels to emit signals with different frequencies.

Because of the combined action of the three gradients, the individual voxels within each slice emit signals that are different in two respects—they have a phase difference in one direction and a frequency difference in the other. Although these signals are emitted at the same time, and picked up by the imaging system as one composite signal at the time of the echo event in each cycle, the reconstruction process can sort the signals into the respective components and display them in the correct image pixel locations.

Image Reconstruction

The next major step in the creation of an MR image is the reconstruction process. Reconstruction is the mathematical process performed by the computer that converts the collected

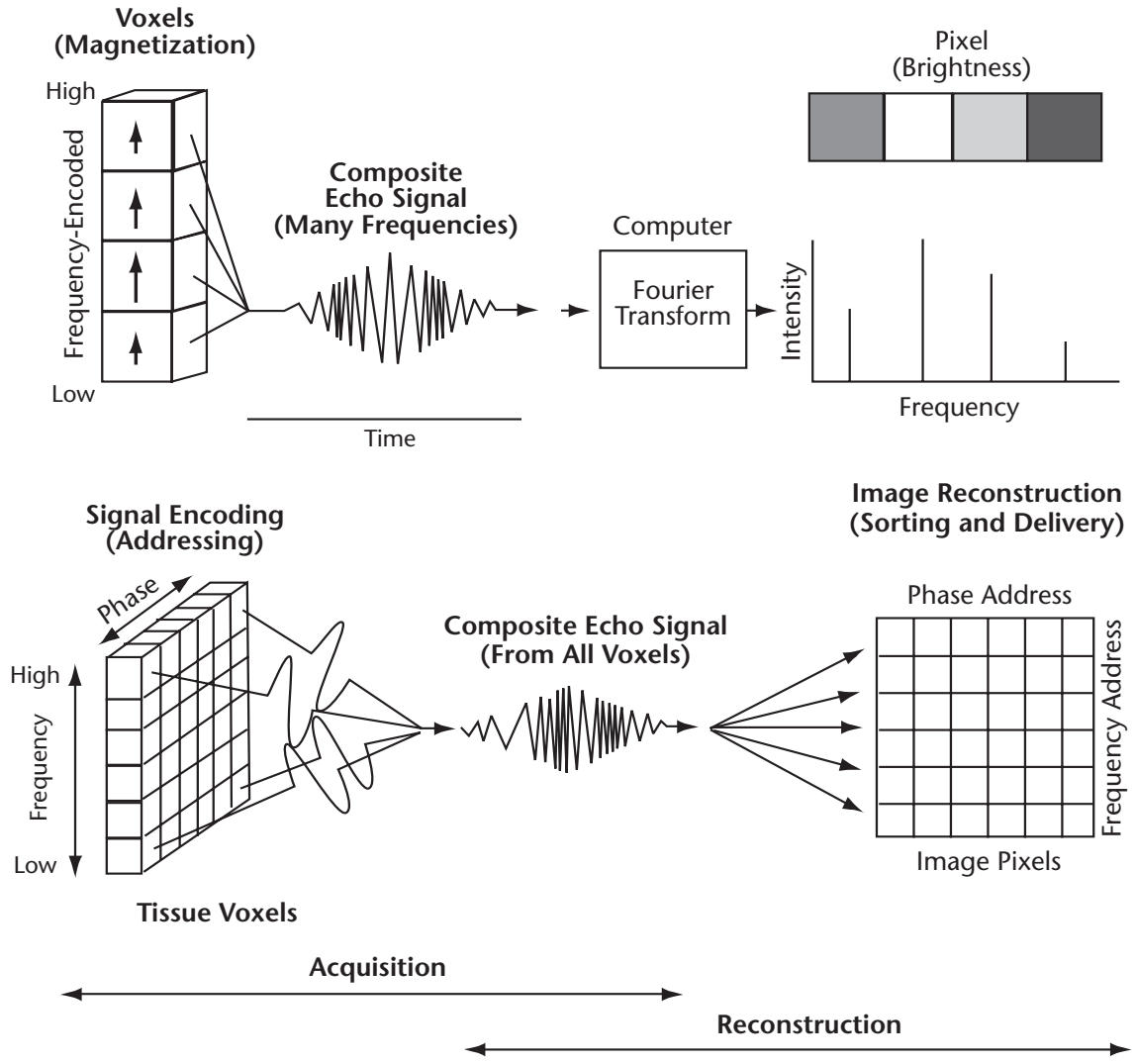


Figure 9-11. The concept of signal encoding (addressing) and image reconstruction (sorting and delivery).

signals in k space into an actual image. There are several reconstruction methods, but the one used for most clinical applications is the 2-D Fourier transformation.

It is a mathematical procedure that can sort a composite signal into individual frequency and phase components. Since each voxel in a row emits a different signal frequency and each voxel in a column a different phase, the Fourier transformation can determine the location of each signal component and direct it to the corresponding pixel.

Let us now use the concept illustrated in Figure 9-11 to summarize the spatial characteristics of the MR image. We will use a postal analogy for this purpose.

In the image each column of pixels has a phase address corresponding to different street names. Each row of pixels has a frequency address corresponding to house numbers. Therefore, each individual pixel has a unique address consisting of a combination of frequency and phase values analogous to a street name and house number.

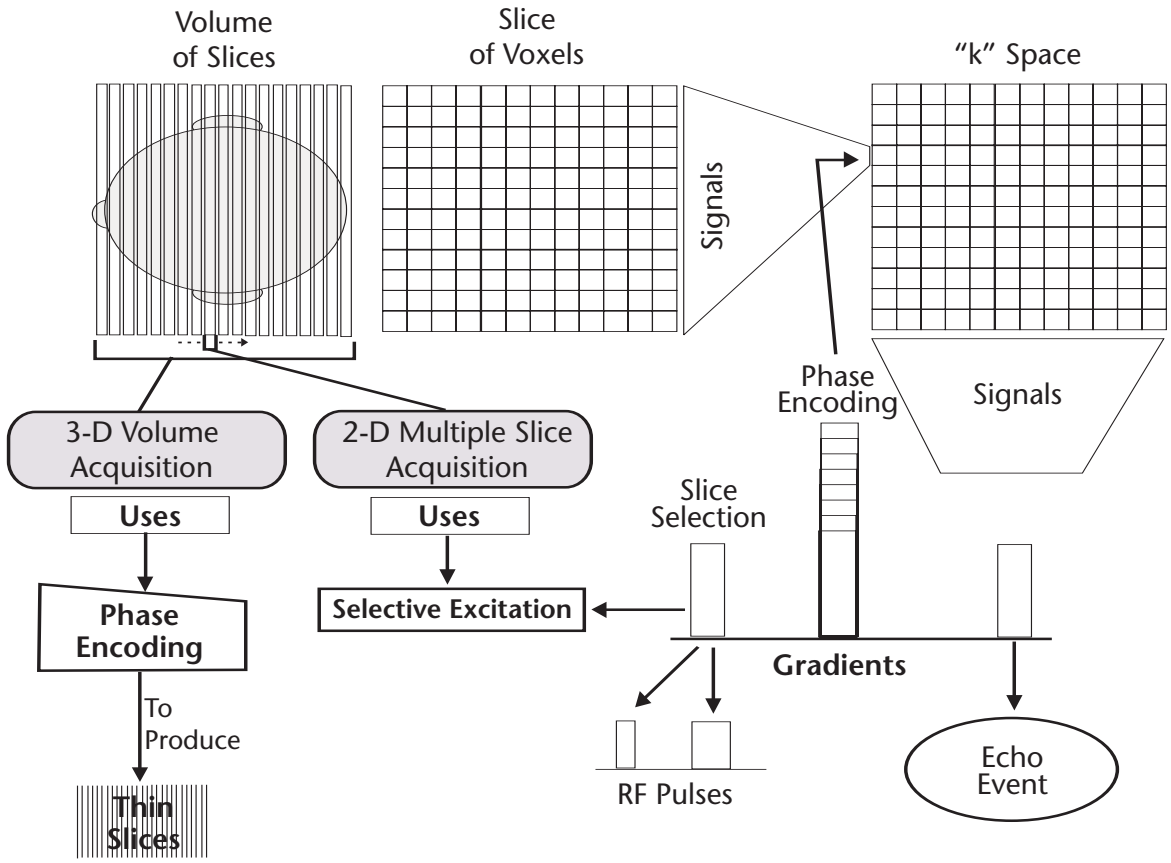
The frequency- and phase-encoding process during acquisition "writes" an address on the signal from each voxel. These signals are mixed together and collected in a "post-office" called k space. The signals ("mail") are then sorted by the Fourier transform process and hopefully delivered to the correct pixel address in the image.

In Chapter 14 we will see that if a voxel of tissue moves during the acquisition process, it might not receive the correct phase address and the signal will be delivered to the wrong pixel. This creates ghost images and streak artifacts in the phase-encoded direction.

The chemical-shift artifact is caused by the difference in signal frequency between tissues containing water and fat. When it is present in an image, signals from the water components and fat will be offset by a few pixels. We will see how this is controlled in Chapter 14.

Mind Map Summary

Spatial Characteristics of the Magnetic Resonance Image



During an MRI procedure a section of a patient's body is first divided into a set of slices, and then each slice is divided into a matrix of voxels. These actions are produced by the gradients.

Two methods can be selected to produce the slices. The most common method, 2-D multiple slice acquisition, applies a gradient so that an individual slice is tuned to a resonant frequency different from the other slice positions. This gradient is turned on when the RF pulses are applied. Therefore, only the tissue in a specific slice is excited and goes through the process to produce signals. An alternate method, 3-D volume acquisition, uses phase-encoding to produce slices. It is generally capable of producing thinner, more contiguous slices.

Two different methods are used to cut a slice into voxels. Phase-encoding is used in one direction, and frequency-encoding in the other. Phase-encoding is produced by applying a gradient to the transverse magnetization during each imaging cycle. To produce sufficient phase-encoding information to permit image reconstruction, many different phase-encoding gradient strengths must be used. In the typical imaging procedure the phase-encoding gradient strength is changed

from cycle to cycle. The strength of the phase-encoding gradient, in effect, directs the signal data into a specific row of k space. All the rows of k space must be filled with data before the image reconstruction can be performed. The number of rows of k space is one of the factors that determine how many imaging cycles must be used, which, in turn, affects image acquisition time.

Frequency-encoding is produced by applying a gradient at the time of the echo event during each cycle.

10

Image Detail and Noise

Introduction And Overview

Two characteristics of the MR image that reduce the visibility of anatomical structures and objects within the body are *blurring* and *visual noise*. These were introduced in Chapter 1 as image quality characteristics. Both image blurring and visual noise are undesirable characteristics that collectively reduce the overall quality of an image and the objects in the image as illustrated in Figures 1-6 and 1-7. In an image, the combined effects of blur and noise produce a "curtain of invisibility" that extends over some objects based on object characteristics. This is shown in Figure 10-1, where we see objects arranged according to two characteristics. In the horizontal direction, the objects are arranged according to size. Decreasing object size corresponds to increasing detail. In the vertical direction, the objects are arranged according to their contrast. The object in the lower left is both large and has a high level of contrast. This is the object that would be most visible under a variety of imaging conditions. The object that is always the most difficult to see is the small, low contrast object, which in Figure 10-1 would be located in the upper right corner.

In every imaging procedure we can assume that some potential objects within the body will not be visible because of the blurring and noise in the image. This loss of visibility is represented by the "curtain" or area of invisibility indicated in Figure 10-1. The location of the boundary

between the visible and invisible objects, often referred to as a *contrast-detail curve*, is determined by the amount of blurring and noise associated with a specific imaging procedure. In general, blurring reduces the visibility of anatomical detail or other small objects that are located in the lower right region. Visual noise reduces the visibility of low contrast objects located in the upper left region.

The imaging protocol determines the boundary of visibility by altering the amount of blurring and noise. These two characteristics are determined by the combination of many adjustable imaging factors. It is a complex process because the factors that affect visibility of detail (blurring) also affect noise, but in the opposite direction. As we will see when a protocol is changed to improve visibility of detail, the noise is increased. Another point to consider is that several of the factors that have an effect on both image detail and noise also affect image acquisition time, which will be discussed in Chapter 11. Therefore, when formulating an imaging protocol one must consider the multiple effects of the imaging factors and then select factor values that provide an appropriate compromise and an optimized acquisition for a specific clinical study with respect to detail (blurring), noise, and acquisition speed.

We will now consider the many factors that have an effect on the characteristics of image detail and noise.

Image Detail

The ability of a magnetic resonance image to show detail is determined primarily by the size of the tissue voxels and corresponding image pixels. Pixel size can be changed without major tradeoffs. However, as we are about to observe, there are significant effects of changing voxel

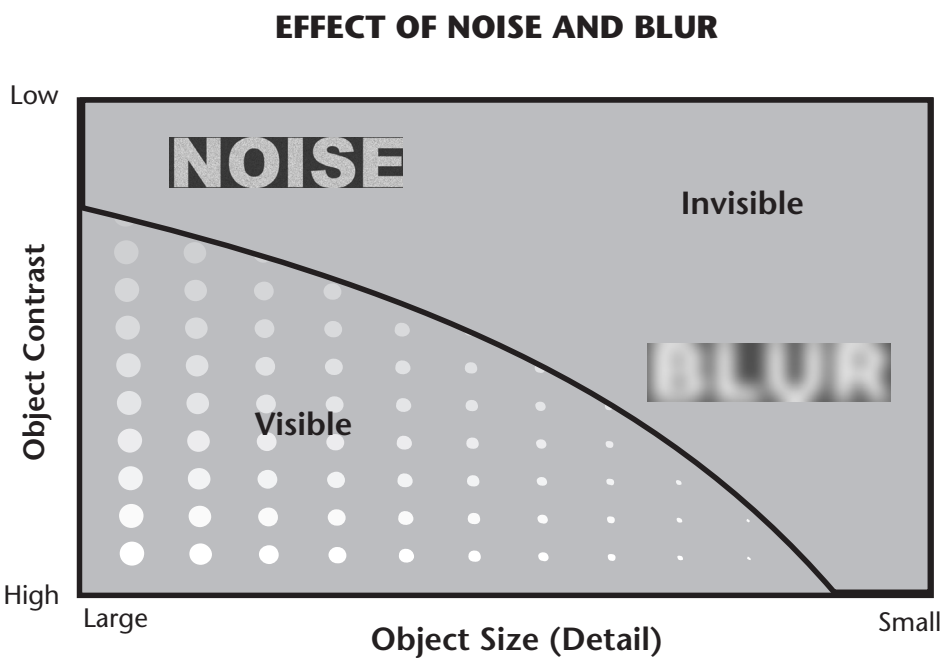


Figure 10-1. The impact of image noise and blurring on object visibility. Noise reduces visibility of low contrast objects. Blur reduces visibility of small objects.

size that must be considered. The real challenge is selecting a voxel size that is optimum for a specific clinical procedure.

In principle, all structures within an individual voxel are blurred together and represented by the signal intensity representing that voxel. It is not possible to see details within a voxel, just the voxel itself. When we view an MR image, we are actually looking at an image of a matrix, or array, of the voxels. We usually do not see the individual voxels because they are so small and they might be interpolated into even smaller image pixels. However, even if we do not see the individual voxels, their size determines the anatomical detail that we can see. The amount of image blurring is determined by the dimensions of the individual voxels.

Three basic imaging factors determine the dimensions of a tissue voxel, as illustrated in Figure 10-2. The dimension of a voxel in the plan of the image is determined by the ratio of the field of view (FOV) and the size of the matrix. Both of these factors can be used to adjust image detail. The thickness of the slice is a factor in voxel signal intensity.

The selection of the FOV is determined primarily by the size of the body part being imaged. One problem that can occur is the appearance of *foldover artifacts* when the FOV is smaller than the actual body section. However, there are artifact suppression techniques that can be used to reduce this foldover problem, as described in Chapter 14. The maximum useful FOV is usually limited by the dimensions and characteristics of the RF coil. The important thing to remember is that smaller image FOVs and smaller voxels produce better visibility of detail.

Matrix size refers to the number of voxels in the rows or columns of the matrix. The matrix size is a protocol factor selected by the

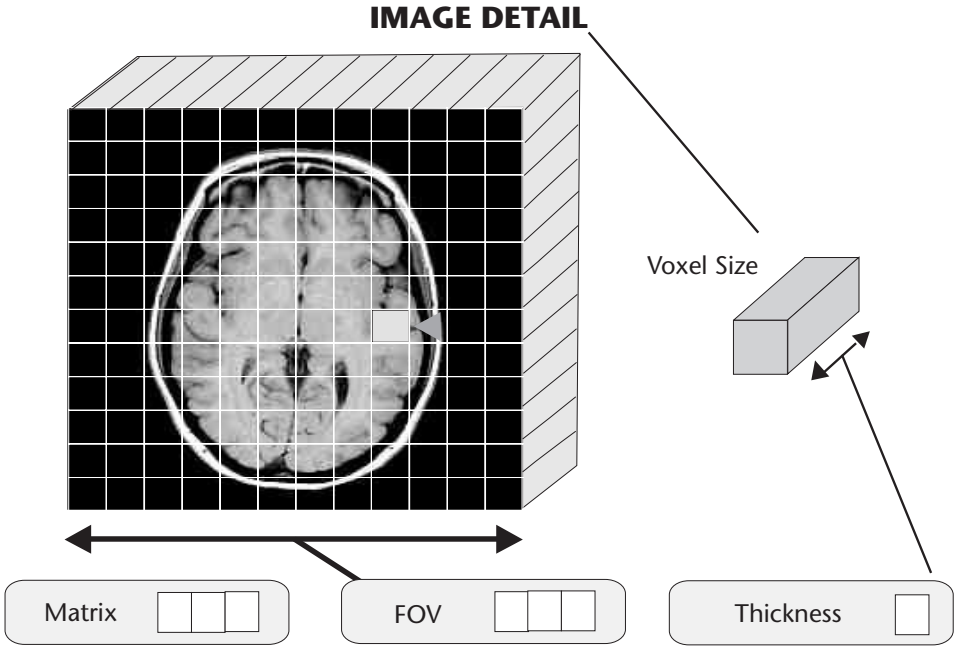


Figure 10-2. Voxel size and detail in MR images is determined by the values selected for the three protocol factors: FOV, matrix size, and slice thickness.

operator before the imaging procedure. Typical matrix dimensions are in the range of 128 to 512 mm.

Noise Sources

Random RF energy can be generated by thermal activity within electrical conductors and circuit components of the receiving system. In principle, the patient's body is a component of the RF receiving system. Because of its mass, it becomes the most significant source of image noise in most imaging procedures. The specific noise source is the tissue contained within the sensitive FOV of the RF receiver coils. Some noise might be generated within the receiver coils or other electronics, but it is usually much less than the noise from the patient's body.

Many devices in the environment produce RF noise or signals that can interfere with MRI. These include radio and TV transmitters, electrosurgery units, fluorescent lights, and computing equipment. All MR units are installed with an RF shield, as described in Chapter 2, to reduce the interference from these external sources. External interference is not usually a problem with a properly shielded unit. When it does occur, it generally appears as an image artifact rather than the conventional random noise pattern.

Signal-To-Noise Considerations

Image quality is not dependent on the absolute intensity of the noise energy but rather the amount of noise energy in relation to the image signal intensity. Image quality increases in proportion to the signal-to-noise ratio. When the intensity of the RF noise is low in proportion to the intensity of the image signal, the noise has a low visibility. In situations where the signals are relatively weak, the noise becomes much more visible. The principle is essentially the same as with conventional TV reception. When a strong signal is received, image noise (snow) is generally not visible; when one attempts to tune in to a weak TV signal from a distant station, the noise (noise) becomes significant.

In MRI, the loss of visibility resulting from the noise can be reduced by either *reducing the noise intensity* or *increasing the intensity of the signals*. This is illustrated in Figure 10-3. Let us now see how this can be achieved.

Voxel Size

One of the major factors that affects signal strength is the volume of the individual voxels. The signal intensity is proportional to the total number of protons contained within a voxel. Large voxels, that contain more protons, emit stronger signals and result in less image noise. Unfortunately, as we have just discovered, large voxels reduce image detail. Therefore, when the factors for an imaging procedure are being selected, this compromise between signal-to-noise ratio and image detail must be considered. The major reason for imaging relatively thick slices is to increase the voxel signal intensity and it also allows shorter TE values.

Field Strength

The strength of the RF signal from an individual voxel generally increases in proportion to the square of the magnetic field strength. However, the amount of noise picked up from the patient's body often increases with field strength because of adjustments in the bandwidth factor for the higher fields. This is described in Chapter 14. Because of differences in system design, no one precise relationship between signal-to-noise ratio and magnetic

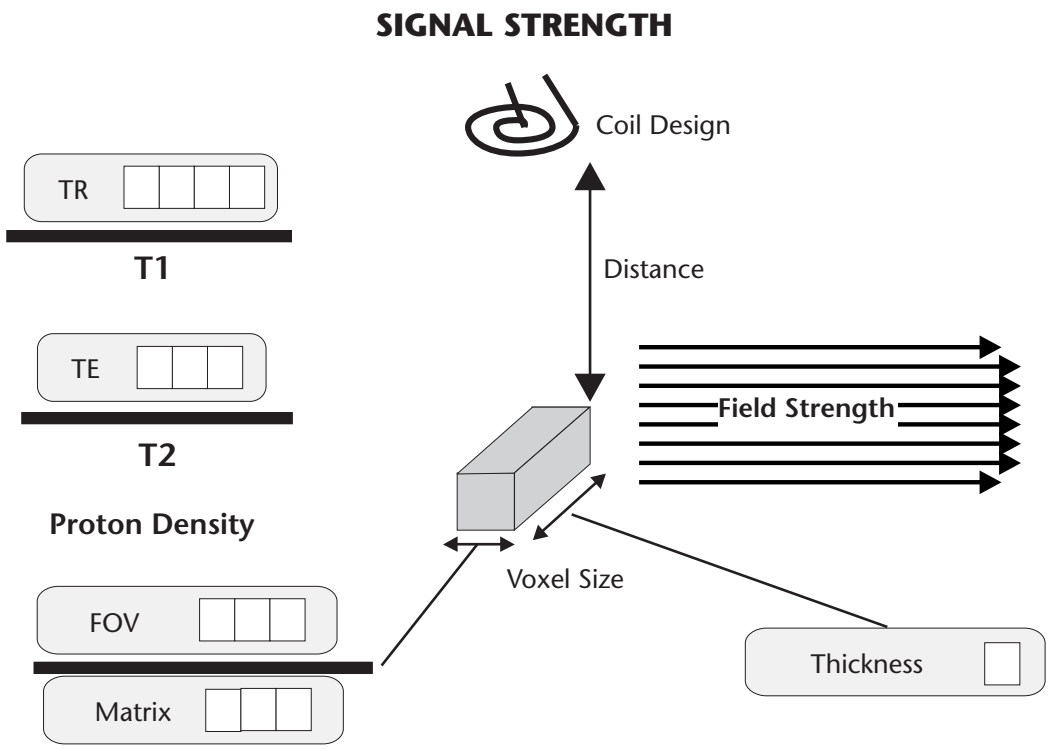


Figure 10-3. Factors that affect the signal-to-noise ratios in MR images.

field strength applies to all systems. In general, MRI systems operating at relatively high field strengths produce images with higher signal-to-noise ratios than images produced at lower field strengths, when all other factors are equal.

Tissue Characteristics

Signal intensity, and the signal-to-noise ratio, depend to some extent on the magnetic characteristics of the tissue being imaged. For a specific set of imaging factors, the tissue characteristics that enhance the signal-to-noise relationship are high magnetic nuclei (proton) concentration, short T1, and long T2. The primary limitation in imaging nuclei other than hydrogen (protons) is the low tissue concentration and the resulting low signal intensity.

TR and TE

Repetition time (TR) and echo time (TE) are the factors used to control contrast in most imaging methods. We have observed that these two factors also control signal intensity. This must be taken into consideration when selecting the factors for a specific imaging procedure.

When a short TR is used to obtain a T1-weighted image, the longitudinal magnetization does not have the opportunity to approach its maximum and produce high signal intensity. In this case, some signal strength must be sacrificed to gain a specific type of image contrast. Also, when TR is reduced to decrease image acquisition time, image noise can become the limiting factor.

When long TE values are used, the transverse magnetization and the resulting signal it produces can decay to very low values. This causes the images to display more noise.

RF Coils

The most direct control over the amount of noise energy picked from the patient's body is achieved by selecting appropriate characteristics of the RF receiver coil. In principle, noise is reduced by decreasing the amount of tissue within the sensitive region of the coil. Most imaging systems are equipped with interchangeable coils. These include a body coil, a head coil, and a set of surface coils as shown in Figure 10-4. The body coil is the largest coil and usually contains a major part of the patient's tissue within its sensitive region. Therefore, body coils pick up the greatest amount of noise. Also, the distance between the coil and the tissue voxels is greater than in other types of coils. This reduces the intensity of the signals actually received by the coil. Because of this combination of low signal intensity and higher noise pickup, body coils generally produce a lower signal-to-noise ratio than the other coil types.

In comparison to body coils, head coils are both closer to the imaged tissue and generally contain a smaller total volume of tissue within their sensitive region. Because of the increased signal-to-noise characteristic of head coils, relatively small voxels can be used to obtain a better image detail.

The surface coil provides the highest signal-to-noise ratio of the three coil types. Because of its small size, it has a limited sensitive region

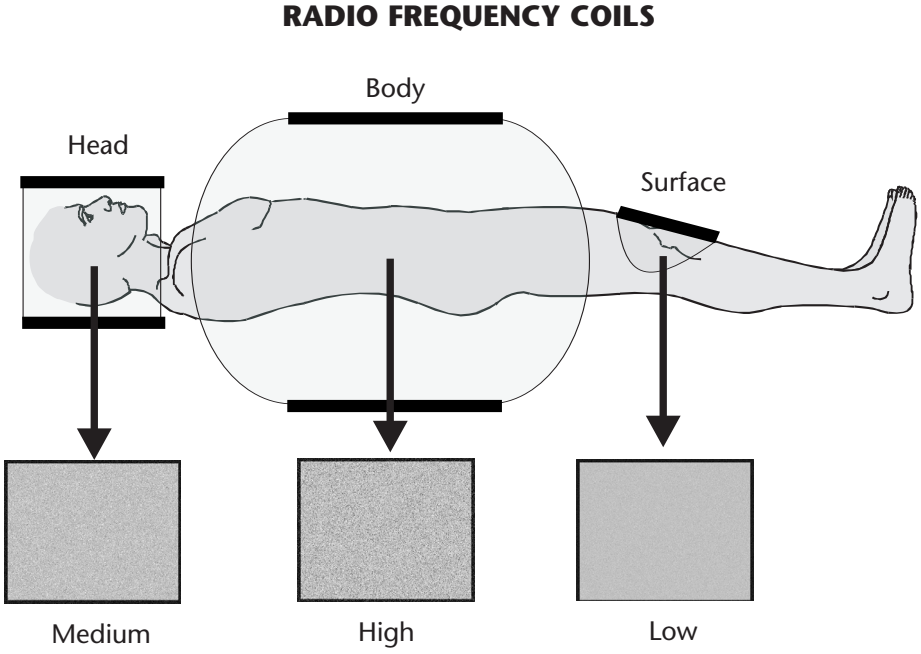


Figure 10-4. Both the amount of noise and the intensity of the signal received depend on the RF receiving coils. The body coil picks up the most noise and the weakest signal, resulting in the highest noise level in the image.

and picks up less noise from the tissue. When it is placed on or near the surface of the patient, it is usually quite close to the voxels and picks up a stronger signal than the other coil types. The compromise with surface coils is that their limited sensitive region restricts the useful FOV, and the sensitivity of the coil is not uniform within the imaged area. This non-uniformity results in very intense signals from tissue near the surface and a significant decrease in signal intensity with increasing depth. The relatively high signal-to-noise ratio obtained with surface coils can be traded for increased image detail by using smaller voxels.

Receiver Bandwidth

Bandwidth is the range of frequencies (RF) that the receiver is set to receive and is an adjustable protocol factor. It has a significant effect on the amount of noise picked up. This is because the noise is distributed over a wide range of frequencies, whereas the signal is confined to a relatively narrow frequency range. Therefore, when the bandwidth is increased, more noise enters the receiver. The obvious question is: Why increase bandwidth? One reason is that a wider bandwidth reduces the chemical shift artifact that will be described in Chapter 14. Also, wider bandwidths are the result of short signal sampling, or "picture snapping" times that are useful in some applications.

Averaging

One of the most direct methods used to control the signal-to-noise characteristics of MR images is the process of averaging two or more

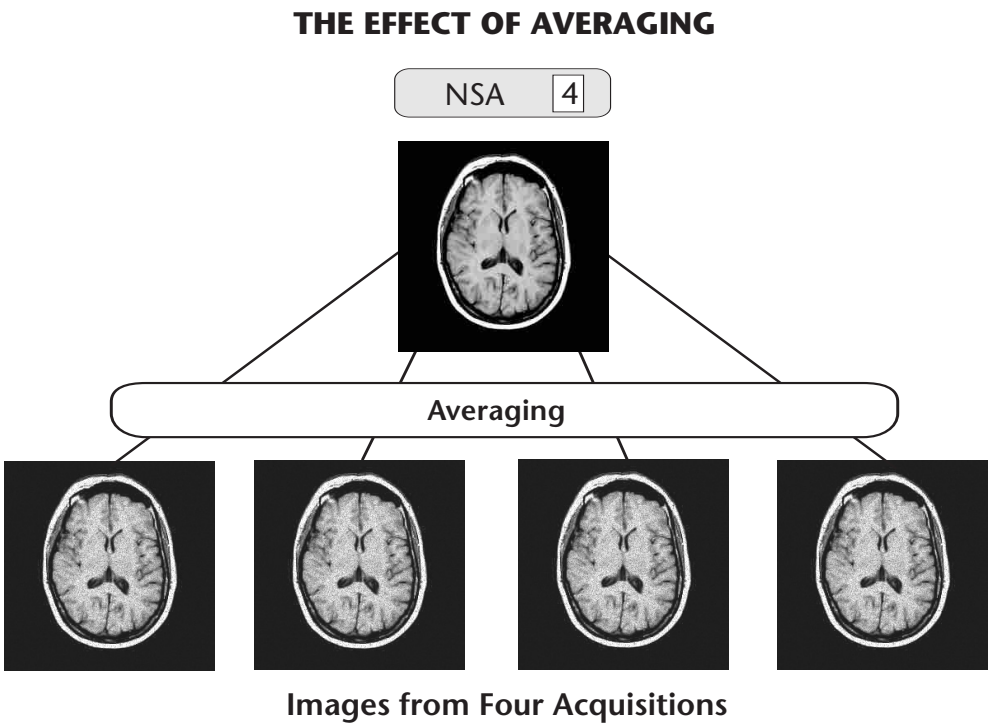
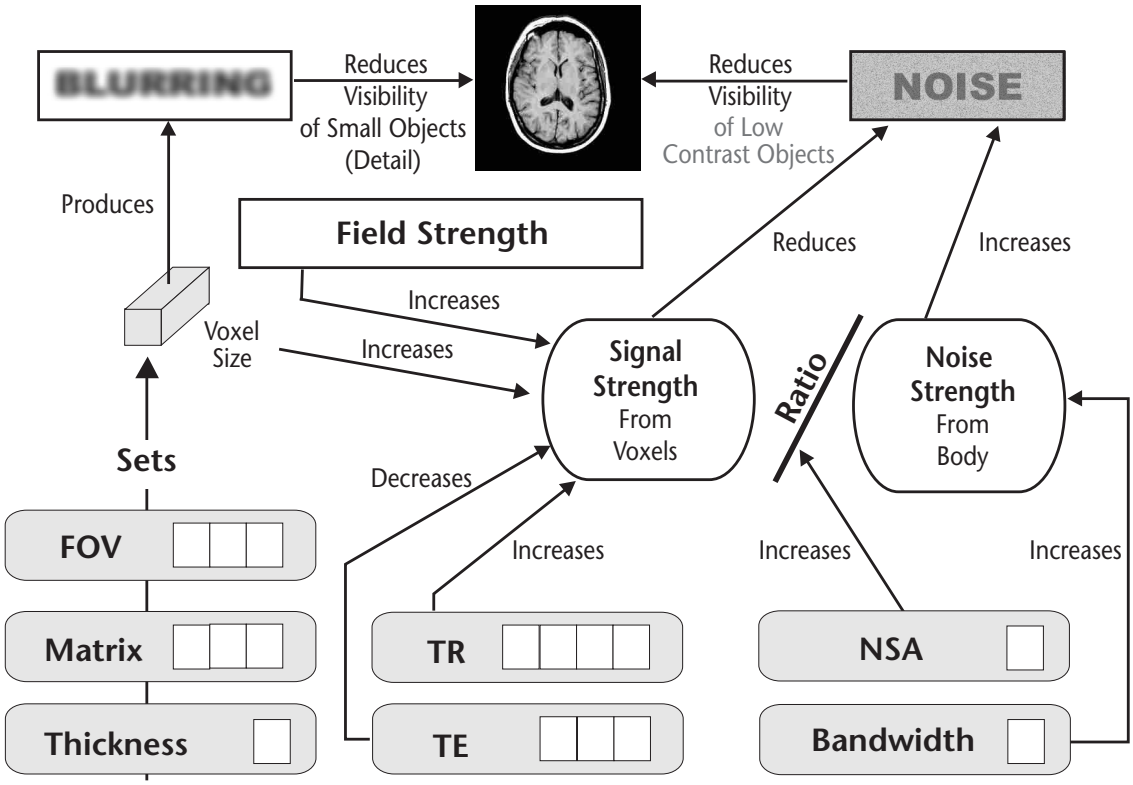


Figure 10-5. An image with reduced noise is created by averaging the signals from four acquisitions.

signal acquisitions. In principle, each basic imaging cycle (phase-encoding step) is repeated several times and the resulting signals are averaged to form the final image as illustrated in Figure 10-6. The averaging process tends to reduce the noise level because of its statistical fluctuation nature, from one cycle to another. You can think of it as acquiring four images by repeating the basic acquisition four times. Then the signal intensities in each pixel position in the four images are averaged to produce an intensity value for the new averaged image.

The disadvantage of averaging is that it increases the total image acquisition time in proportion to the number of cycle repetitions or number of signals averaged (NSA). The NSA is one of the protocol factors set by the operator. Typical values are 1 (no averaging), 2, or 4, depending on the amount of noise reduction required. The general relationship is that the NSA must be increased by a factor of 4 to improve the signal-to-noise ratio by a factor of 2. The signal-to-noise ratio is proportional to the square root of the NSA. Sometimes the noise contribution from independent acquisitions adds; sometimes it cancels. Since the signals always add, adding or averaging independently acquired images improves the signal-to-noise ratio.

Mind Map Summary
Image Detail and Noise



Two important image quality characteristics are blurring, which reduces visibility of small objects or detail, and image noise, which reduces visibility of low contrast objects. Both of these characteristics depend on design characteristics of the imaging system and the combination of selected protocol factors. The principal source of blurring in an MR image is the voxel size. This is because all tissues within an individual voxel are blurred together and represented by one signal. An image does not display any detail within the individual voxels. It is just a display of a matrix of voxels. Voxel size, and the resulting blurring, can be adjusted with the three protocol factors: FOV, matrix, and slice thickness.

The level of noise that appears in an image depends on the relationship (ratio) of the signal strength from the individual voxels and the noise strength coming from a region of the patient's body. The visible noise is reduced by increasing signal strength. This can be done by increasing the magnetic field strength, increasing voxel size, increasing TR, and decreasing TE. The field strength is a design characteristic and cannot be changed by the operator. Increasing voxel size to decrease noise has the adverse effect of also increasing blurring. Voxel sizes must be chosen to provide an appropriate balance between blurring and noise.

The noise strength picked up from the patient's body is determined by the mass of tissue contained within the sensitive pickup region of the RF coils. Surface coils that cover a relatively small anatomical region and are also close to the signal source (voxels) produce a high signal-to-noise relationship that results in lower image noise. The RF receiver bandwidth can be adjusted to block some of the noise energy from being received. However, decreasing the bandwidth to reduce noise has the adverse effect of increasing the chemical shift artifact.

Signal averaging is a useful technique for reducing noise but has the adverse effect of increasing acquisition time.

11

Acquisition Time And Procedure Optimization

Introduction And Overview

A significant time is usually required for the acquisition of an MR image or a set of multi-slice images. It is generally desirable to keep acquisition time as short as possible for a variety of reasons including: increasing patient throughput, reducing problems from patient motion, and performing dynamic imaging procedures. It is technically possible to reduce acquisition time by changing several protocol factors such as TR and matrix size. However, the basic protocol factors that affect acquisition time also have an effect on the characteristics and quality of the image. In general, changing factors to reduce acquisition time also decreases image quality.

When setting up an imaging protocol the acquisition time must be considered with respect to the necessary requirements for image quality. An optimized protocol is one in which there is a good balance among the image quality requirements and acquisition time.

In this chapter we will consider the various factors that determine acquisition time, how they relate to image characteristics and quality, and how the various requirements can be balanced.

Acquisition Time

Figure 11-1 illustrates a basic MR image acquisition process and identifies the factors that determine acquisition time. We recall that an

image acquisition consists of an imaging cycle that is repeated many times. The duration of the cycle is TR, the protocol factor that is used to control certain image characteristics such as contrast. The primary factor that determines the number of times the cycle must be repeated is the number of rows of k space that must be filled. This, in turn, is determined by the image matrix size in the phase-encoded direction. As we observed in Chapter 9, it is the strength of the phase-encoding gradient during each cycle that directs the signals into a specific row of k space. Therefore, the phase-encoding gradient must be stepped through a range of different strengths, with the number of steps corresponding to the number of rows of k space that must be filled. The number of rows of k space that must be filled is determined by the image matrix size in the phase-encoded direction. In Chapter 10 we saw that matrix size is one of the protocol factors that determines image detail and also has an effect on image noise. Reducing matrix size in an effort to reduce acquisition time without changing the FOV will decrease image detail.

In a basic acquisition the time required is TR multiplied by the matrix size in the phase-encoded direction.

$$\text{Time} = \text{TR} \times \text{matrix size}$$

In an acquisition, the time can be adjusted by changing either of these two factors. However, we will soon learn that there are some additional factors that are related to certain imaging

ACQUISITION TIME

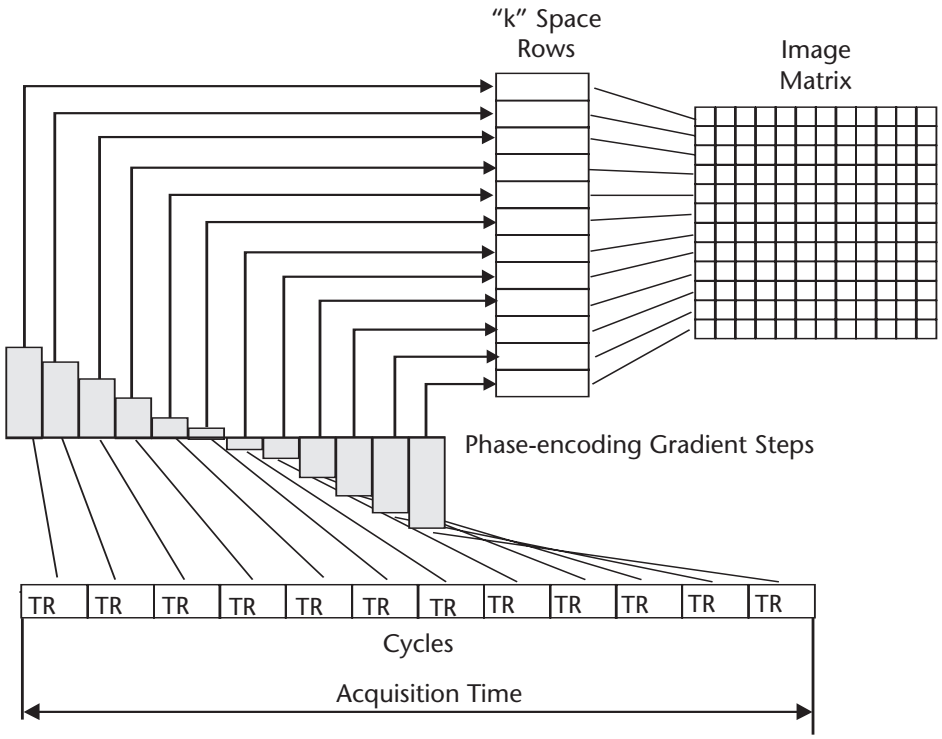


Figure 11-1. The acquisition time is determined by the cycle duration, TR, and number of cycles required to fill the k space.

techniques and methods that also have an effect on acquisition time. When applied, some will increase while others will decrease the time.

Cycle Repetition Time, TR

Until this point, we have used TR as an adjustable protocol factor to control the type of image (T1, T2, PD) that is being acquired. We recall from Chapter 6 that with spin echo imaging TR values of approximately 500 msec are used to acquire T1 images, but values as long as 2000 msec are required for PD and T2 images. TR is the time that is required, within an individual cycle, for the longitudinal magnetization to relax and recover to the appropriate level for the type of image that is being acquired. When the inversion recovery method is used, especially for fluid suppression (Chapter 8), even longer TR values are required. The point is that certain TR values are required to produce specific image types and they cannot be arbitrarily shortened to reduce acquisition time. As we learned in Chapter 7, much shorter TR values can be used with the gradient echo methods. This is why gradient echo methods are generally much faster than the spin echo methods but are not appropriate for many imaging procedures.

Matrix Size

We recall from Chapter 10 that the matrix size and the FOV are the two factors that determine voxel size in the plane of the image. Voxel size determines the amount of blurring and image detail. Small voxels and minimum blurring are required for good detail. If the matrix size is reduced without changing the FOV, voxel size will be increased and there will be a reduction in image detail. Let us now use Figure 11-2 to see how matrix size can be adjusted to obtain an optimum balance between acquisition time and image detail.

It is only the matrix size in the phase-encoded direction that has an effect on acquisition time. Recall that this is because this matrix dimension determines the number of rows of k space that must be filled. Matrix size in the frequency-encoded direction does not affect acquisition time.

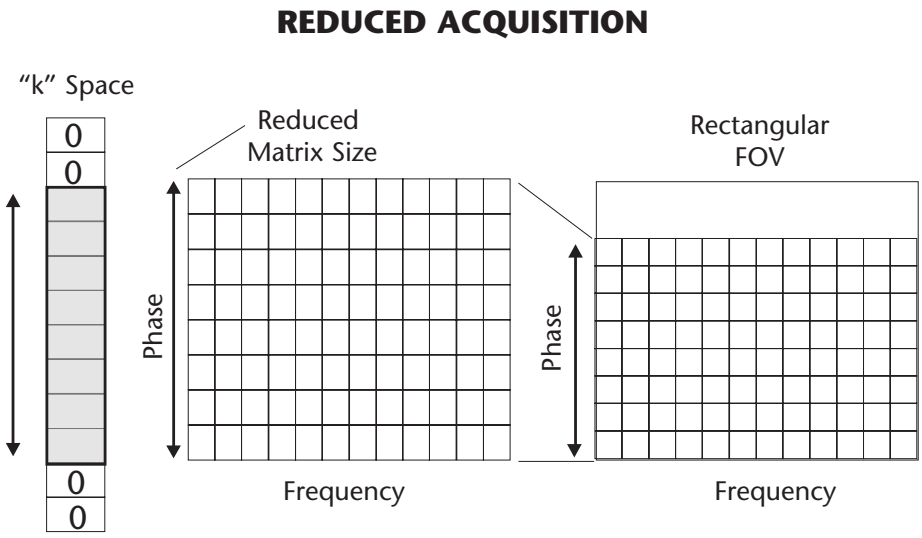


Figure 11-2. Reducing matrix size in the phase-encoded direction reduces the lines of k space to be filled. Then, reducing the FOV to a rectangle reduces the voxel size and restores image detail.

Reduced Matrix in Phase-Encoded Direction

One approach to optimizing an acquisition is to reduce the matrix size in the phase-encoded direction to a value that is less than the matrix size in the frequency-encoded direction. It is a common practice to set the matrix size in the phase-encoded direction to some percentage (such as 80%) of the size in the frequency-encoded direction. This has the effect of reducing acquisition time by 20% with minimal effect on image detail.

The selectable basic matrix sizes are binary multiplies such as 128, 256, 512, and 1024. This is necessary in order for the k space data to be in a form that can be easily processed in the image reconstruction phase. When basic matrix size is selected, it is one of these values, with 256 being the most common for most procedures. When the matrix size is reduced by some percentage in the phase-encoded direction, the computer fills in the unused row of k space with zeros to make the reconstruction process work.

Rectangular Field of View

Decreasing matrix size without changing the FOV does produce an increased voxel size. However, if the FOV can be reduced in the phase-encoded direction, the voxel size will be decreased and image detail will be maintained. The use of a rectangular FOV is a way of optimizing acquisition time and image detail if a rectangular FOV works for the specific anatomical region that is being imaged.

By combining a reduced matrix size (in the phase-encoded direction) with a rectangular FOV, acquisition time can be reduced without affecting image quality. This is one step in optimizing a procedure.

Half Acquisition

One way of filling k space with a reduced number of acquisition cycles is illustrated in Figure 11-3. In addition to half acquisition, this method might be referred to by names such as *half scan* and *half Fourier*.

This method is based on the fact that in a usual acquisition there is symmetry between the two halves of k space. This occurs because of the way the phase-encoding gradients are stepped as illustrated in Figure 11-1. Generally, during the first half of the acquisition the gradient is applied with a high strength for the first cycle. It is then stepped down from cycle to cycle reaching a very low, or zero, value near the center of the acquisition. During the second half of the acquisition the gradient strength is stepped back up to the maximum value for the last imaging cycle. In principle, the phase-encoding steps in the second half are a mirror image of the steps in the first half. Now let us return to Figure 11-3. When using the half acquisition method only the first half of k space is filled directly. Then, the data that was acquired during the first half is mathematically "flipped" and used to fill the second half of k space. This makes it possible to fill all of the rows of k space in approximately half of the normal acquisition time. The actual acquisition time will be slightly more than half because there must be some overlap in the data to make this process work. This is a method that can be used to reduce acquisition time. However, it results in an increase in image noise because the image is being formed with a smaller number of acquired signals. It has the opposite effect of using the technique of averaging to reduce noise that was introduced in Chapter 10. With respect to image noise, using half acquisition is equivalent to setting the NSA to a value of one half.

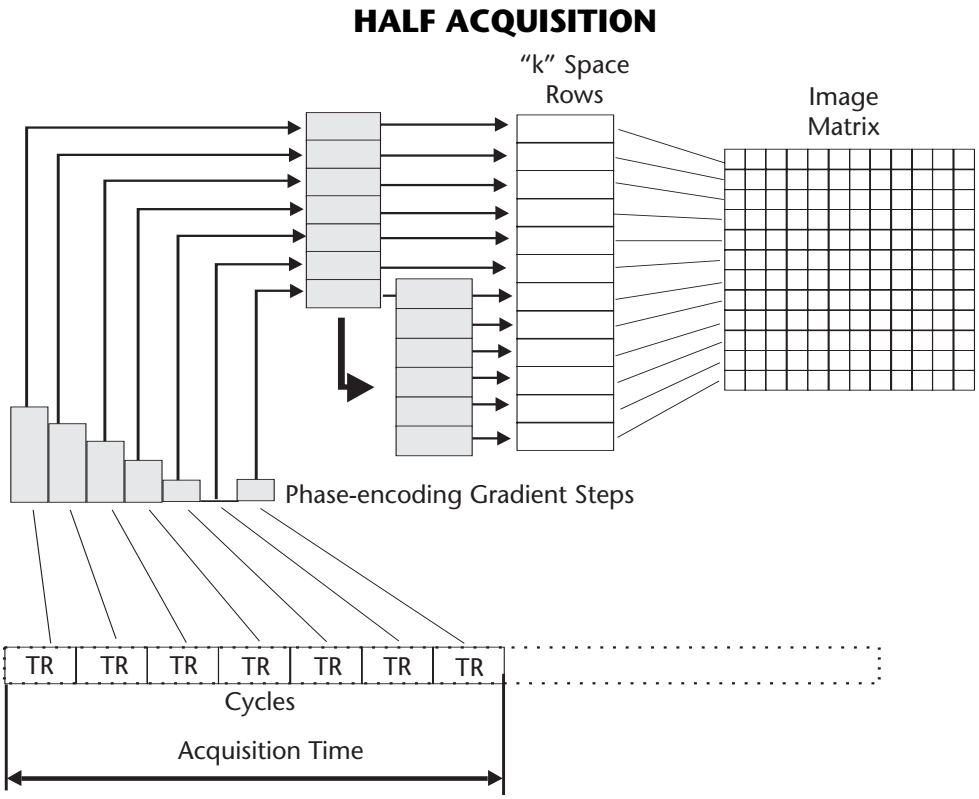


Figure 11-3. Acquisition time is reduced by using half acquisition and duplicating the data to fill the second half of k space.

Signal Averaging

We were introduced to the process of signal averaging in Chapter 10 as a technique that is used to reduce image noise. We now return to signal averaging to see how it affects acquisition time and how it can be combined with other factors to optimize an acquisition procedure. In Figure 11-4 we first see the general relationship between the image acquisition cycles and the phase-encoding gradient steps in a basic acquisition. This is an acquisition where neither signal averaging nor any of the fast imaging methods to be discussed later are used. In this basic acquisition one imaging cycle, with duration of TR, is used for each phase-encoding step. Therefore, the number of acquisition cycles is equal to the number of lines in k space that must be filled, which is equal to the image matrix size in the phase-encoded direction. We will now see that there are acquisition techniques in which there is not this one-to-one relationship between the acquisition cycles and the phase-encoding steps.

The first of these is the use of signal averaging, which is also illustrated in Figure 11-4. The averaging is achieved by repeating the cycle several times for each phase-encoding step. This process does not change the number of phase-encoding steps required, but it does increase the number of cycles in the acquisition. In the example shown, the NSA protocol factor is set to 4. The cycle for each phase-encoding step is repeated four times and the total acquisition

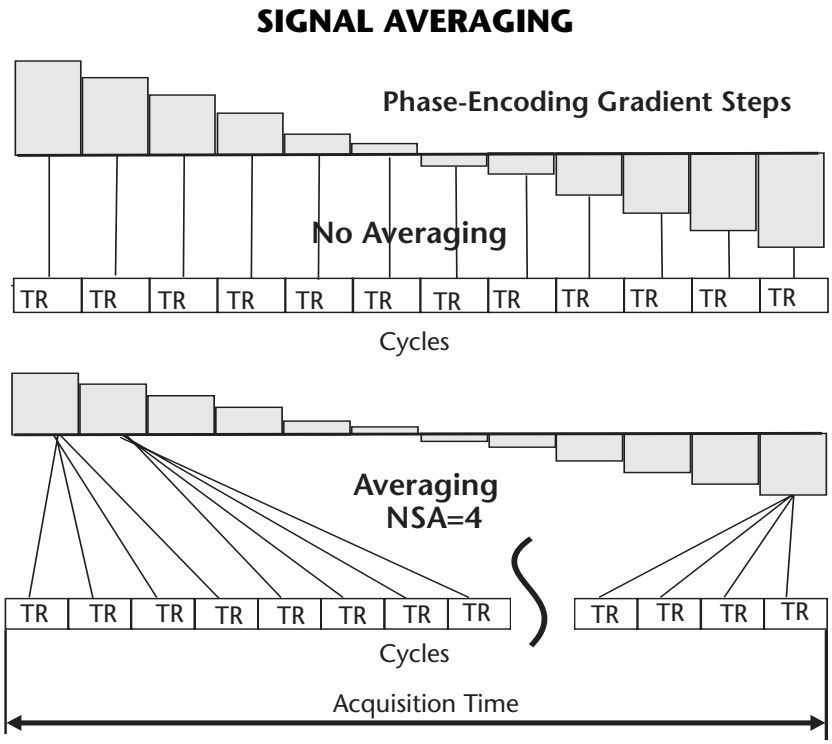


Figure 11-4. Acquisition time is increased by the factor NSA when averaging is used.

time is increased by a factor of 4. When averaging is used, the acquisition time becomes:

$$\text{Time} = \text{TR} \times \text{matrix size} \times \text{NSA}.$$

Protocol Factor Interactions

Up to this point we have observed that each of the protocol factors that affect acquisition time (TR, matrix size, and NSA) also have an effect on the image quality characteristics. The relationship becomes somewhat complex because some factors affect more than one characteristic. A good example is the factors that determine voxel size. As we have seen in Chapter 10 small voxels produce high image detail but result in higher image noise.

In Figure 11-5 we consider three important imaging goals that are affected by some of the same protocol factors. These goals are high image detail, low image noise, and acquisition speed. We have placed these goals at the three corners of a triangle. We can think of this as a type of ball field with three goals. The objective is to move the ball closer to the goals. However, as we move the ball in the direction of one goal, we are moving away from the other goals. In this analogy the location of the ball represents the operating point of a specific acquisition protocol and is determined by the combination of protocol factors used. An optimized protocol is one in which the selected factors produce the best balance among high detail, low noise, and speed for a specific clinical procedure. In some procedures, high detail might be the most important characteristic, but it is obtained at some sacrifice of achieving low noise. The noise can then be reduced by averaging but at the cost of an increased acquisition time.

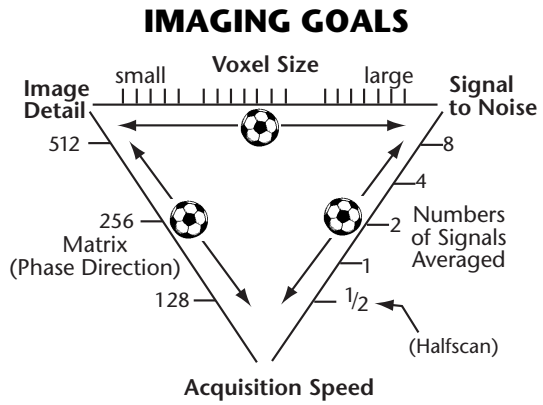


Figure 11-5. Several imaging goals must be considered together when selecting protocol factors.

Developing An Optimized Protocol

An acquisition protocol can be rather complex because of the large number of factors that must be adjusted and the interaction of many of the factors with different image quality characteristics and acquisition times. One approach to developing a good protocol is to address the specific image characteristics in this order:

- Contrast Sensitivity
- Image Detail
- Spatial Characteristics and Methods
- Image Noise
- Artifact Reduction

Contrast Sensitivity

The first step is to select an imaging method and factors (generally TR and TE) to give good contrast sensitivity and visualization for the specific tissue characteristic (T1, T2, PD) or type of fluid movement that is to be observed.

Also, consider if one of the selective signal suppression techniques (Chapter 8) will be helpful in improving contrast.

Image Detail

The next step is to consider the visibility of anatomical detail that is required for the specific clinical procedure. This should lead to the selection of an appropriate slice thickness, which is both the largest dimension of a voxel and the one that can be adjusted over a considerable range.

FOV is generally determined by the anatomy that is to be covered. A rectangular FOV can be used to permit reduced matrix in the phase-encoded direction without a reduction in detail.

Using a reduced matrix percentage in the phase-encoded direction can optimize the matrix size.

It is important to not go for more detail than is actually needed because, as we have seen, the small voxels cause an increase in image noise.

Spatial Characteristics and Methods

Generally, the 2-D multiple slice acquisition would be used unless thin slices are required for good detail and then the 3-D volume acquisition would be more appropriate.

The number of slices and slice orientations are determined by the anatomical regions that are to be covered and the desired views.

Image Noise

Voxel size (slice thickness, FOV, and matrix) has already been set to give the necessary image detail. TR and TE values have been selected to give the desired contrast characteristics. Also, magnetic field strength is a fixed value for the system being used. All these factors together, along with the anatomical volume of tissue that is to be imaged and the type of RF coil selected for the procedure, establish a basic level of image noise. This level of noise

might or might not be acceptable. If it is not acceptable, averaging can be applied with the NSA parameter set to the minimum value that will give acceptable image quality, keeping in mind that this is increasing acquisition time.

Artifact Reduction

One of the final issues in developing a protocol is to consider the types of artifacts that might occur and then include in the protocol one or more of the artifact reduction techniques that are described in Chapter 14.

Fast Acquisition Methods

Until this point, we have seen that there are three factors that determine acquisition time: TR, matrix size, and NSA. This applies to all of the general methods in which only one line of k space is filled in each imaging cycle. We recall that this is when only one phase-encoding step occurs in each cycle. There are several methods that are capable of filling multiple rows of k space in one cycle. Two of these methods, echo planar and GRASE, were described in Chapter 7. Both of these are gradient echo methods. It is also possible to fill multiple rows of k space with the spin echo imaging methods by using the technique that we are about to describe. When this is done, the methods are generally known as *fast* or *turbo* spin echo.

There are many reasons to want to speed up an image acquisition process. It can increase patient throughput and reduce overall operating cost; shorter procedures are more comfortable for patients; there are less problems with patient motion. Also, some dynamic studies require the rapid acquisition of a series of images. Another advantage of using some of the fast acquisition methods is to compensate for other factors that increase acquisition time. A good example is when using a fluid suppression technique as described in Chapter 8. This method requires very long TR values that would produce extremely long, and impractical, acquisition times. However, by combining it with a fast acquisition method, it becomes a very useful technique. The 3-D volume acquisition method that is used to produce thin slices and high image detail also requires a long acquisition time. This is because multiple repetitions must be made since phase-encoding is used to cut the slices, as described in Chapter 9. This becomes a more practical method when used with a fast acquisition technique.

Figure 11-6 illustrates the fast or turbo technique. We recall from Chapter 6 that it is possible to produce multiple spin echo signals within one cycle by applying several 180° pulses after each 90° excitation pulse. At that time we were using each of the echo signals to produce different images, each with a different TE value. This is the multiple echo method that makes it possible to produce both a PD image (with a short TE) and T2 images with the longer TE values in the same acquisition.

Fast or turbo spin echo uses multiple spin echo signals but for a different purpose. With this technique, a different phase-encoding gradient strength (step) is applied for each of the echo signals. Therefore, each echo signal fills a different row of k space, and multiple rows are filled within each cycle. The result is a reduced acquisition time. The reduction in time depends on the number of echo signals produced in each cycle and is one of the protocol variable factors. This number becomes the *speed factor* (or turbo factor) and acquisition time is now:

$$\text{Time} = \text{TR} \times \text{matrix size} \times \text{NSA/speed factor.}$$

Speed factors vary over a considerable range of possible values. In general, there is some

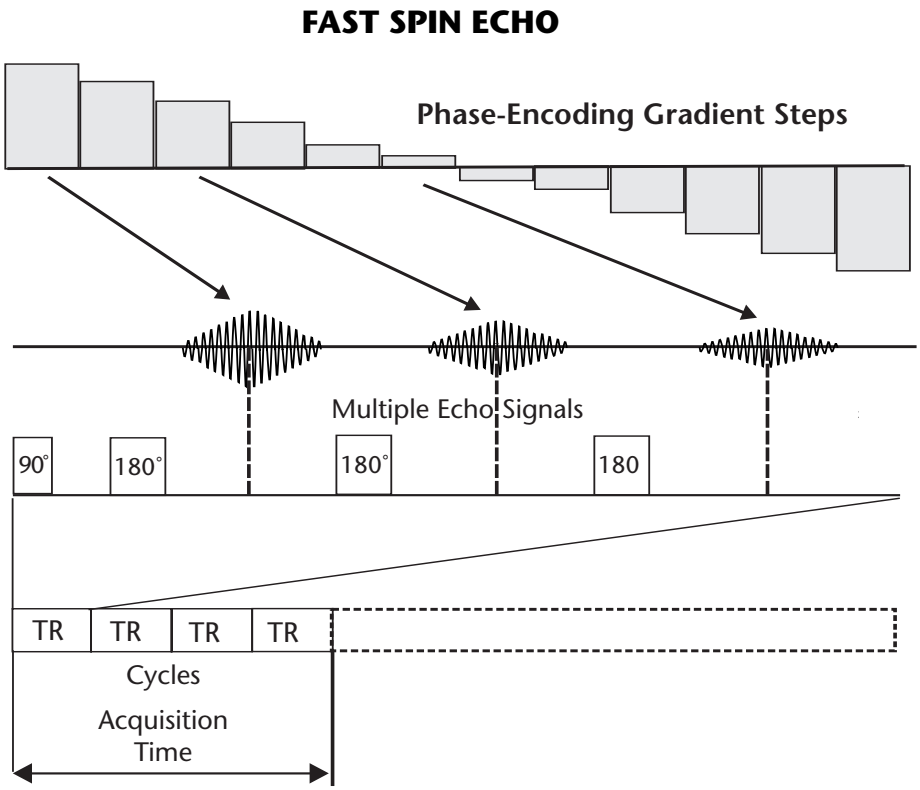


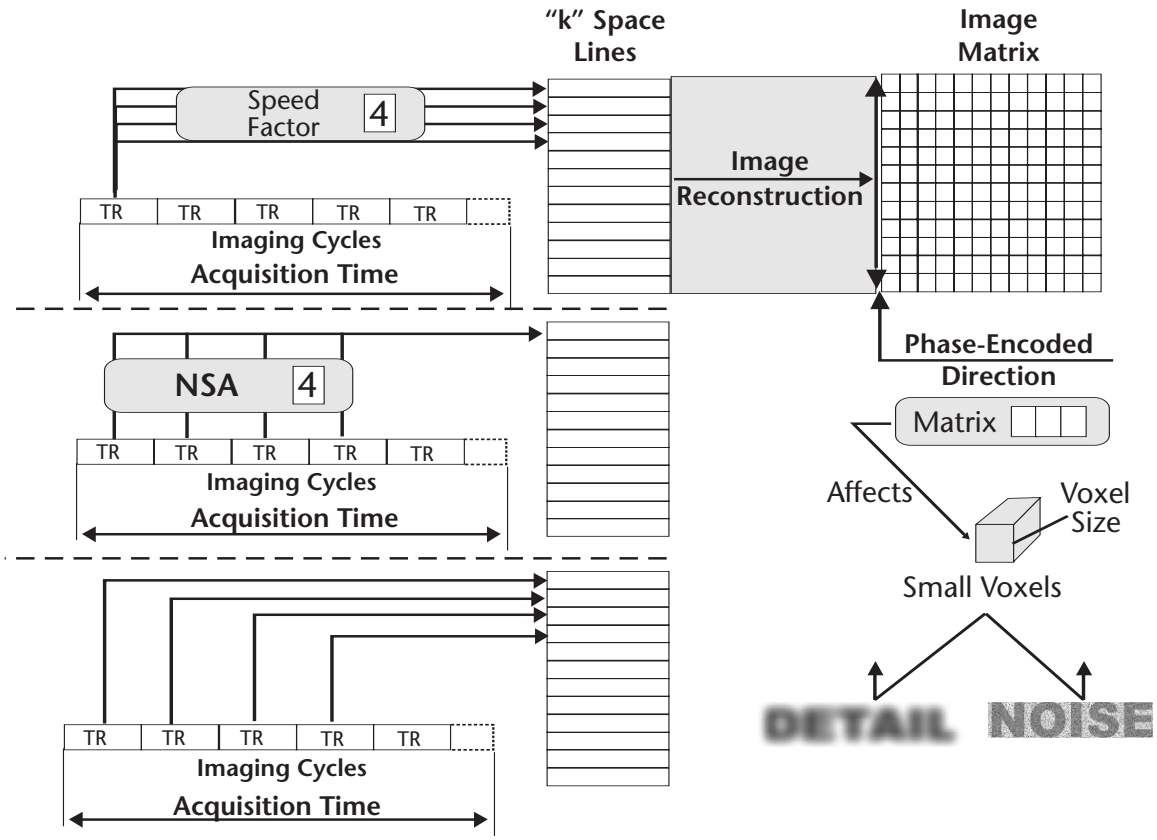
Figure 11-6. The fast spin echo method reduces acquisition time by producing multiple phase-encoded signals within each cycle.

reduction in signal strength (and increased noise) when high speed factor values are used.

With this technique k space is not filled in a progressive bottom-to-top order. The order of filling can be adjusted. One factor to consider is the TE value for the image. Since several echo signals, each with a different TE value, are produced in each cycle and used to form the same image, what is the "correct" TE for the image as far as contrast is concerned? When setting up the protocol for this method, an "effective" TE value is selected. This is the general TE value of the signals that are directed to the more central rows of k space. The data in these rows determine the general contrast characteristics of the image. The data in the outer rows of k space contribute more to the detail characteristics of the image.

Mind Map Summary

Acquisition Time And Procedure Optimization



The MR image acquisition process consists of an imaging cycle (like a heartbeat) that must be repeated many times. The total acquisition time is determined by the duration of each cycle; that is the protocol factor, TR, and the number of cycles required. In a basic imaging procedure, the number of cycles is determined by the number of rows of k space that must be filled, which, in turn, is determined by the image matrix size in the phase-encoded direction. The matrix size can be decreased but this results in an increased voxel size and reduced image detail. When setting up an imaging protocol, attention should be given to selecting a voxel size that provides an appropriate balance between image detail and image noise.

Signal averaging is used to reduce image noise, but it also results in an increased acquisition time because cycles must be repeated according to the selected NSA value.

There are several imaging methods that are capable of filling multiple lines of k space during one imaging cycle. These are some of the fast imaging methods that include fast (turbo) spin echo, echo planar, and GRASE. There is an adjustable speed factor associated with each of these methods.

12

Vascular Imaging

Introduction And Overview

One of the important characteristics of MRI is its ability to create an image of flowing blood and vascular structures without having to inject contrast media. With MRI the contrast between the blood and the adjacent stationary tissue is produced by interactions between the *movement* of the blood and certain events within the imaging process. This is very different from x-ray angiography where the image shows the presence of blood (or contrast media). In MRI the image displays blood-filled vessels, but it is the movement of the flood that actually produces the contrast.

It is a somewhat complex process because under some imaging conditions the flowing blood will produce increased signal intensity and will appear bright, while under other conditions very little or no signal will be produced by the flowing blood and it will be dark. We will designate these two possibilities as "bright blood" imaging and "black blood" imaging, as indicated in Figure 12-1. There are several different physical effects that can produce both bright blood and black blood as indicated. These effects can be divided into three categories:

1. Time effects.
2. Selective saturation.
3. Phase effects.

Under each of these categories there are several specific effects that can produce contrast. Unfortunately, in addition to producing

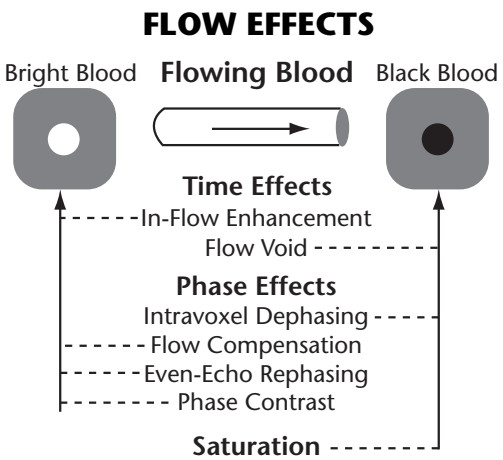


Figure 12-1. The physical effects that produce contrast between flowing blood and the surrounding tissue background.

useful image contrast, the movement of blood often produces undesirable artifacts within the image. These will be considered in Chapter 14.

Some of the flow effects can appear in virtually any image that contains blood vessels. In this chapter we will first explore the various effects associated with flowing blood and show how they can be controlled and used. Then we will apply these effects to specific angiographic procedures.

Time Effects

The time effects are related to the movement of blood during certain time intervals within the acquisition cycle. They are sometimes referred to as the *time-of-flight* effects. The production of bright blood is related to the TR time interval whereas the production of black blood is related to the TE time interval.

Flow-Related Enhancement (Bright Blood)

The process that causes flowing blood to show an increased intensity, or brightness, is illustrated in Figure 12-2. This occurs when the

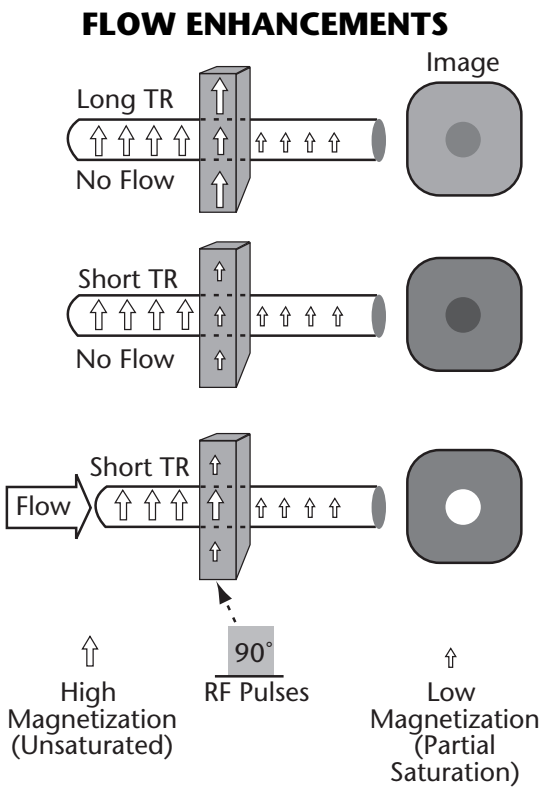


Figure 12-2. Flow-related enhancement produces bright blood because the background tissue is partially saturated by the use of short TR values.

direction of flow is through the slice, as illustrated. It is also known as the *in-flow* effect and is used to produce contrast in in-flow contrast angiography. The degree of enhancement is determined by the relationship of flow velocity to TR. Three conditions are illustrated. The arrow indicates the amount of longitudinal magnetization at the end of each imaging cycle. Because of the slice-selection gradient the RF pulses affect only the blood within the slice.

Let us start by considering using a long TR in the absence of flow in the vessel. The longitudinal magnetization regrows to a relatively high value during each cycle, as indicated at the top of the figure. This condition produces a relatively bright image of both the blood and

the stationary tissue and little contrast between the two. With this as a reference condition, let us now see what happens when a short TR value is used. Each cycle will begin before the longitudinal magnetization has approached its maximum. This results in reduced signal intensity and a relatively dark image because both the non-flowing blood and tissue remain partially saturated.

The next step is to see what happens if the blood is flowing into the slice. The effect of flow is to replace some of the blood in the slice with fully magnetized blood from outside the slice. The increased magnetization at the end of each cycle increases image brightness of the flowing blood. The enhancement increases with flow until the flow velocity becomes equal to the slice thickness divided by TR. This represents full replacement and maximum enhancement and brightness.

In order for blood to appear bright, the surrounding background stationary tissue must be relatively dark. Therefore, a part of any bright blood imaging process is to suppress the signals from the surrounding tissue. With the in-flow effect, the background tissue is dark because it is partially saturated from the use of the short TR values. When long TR values are used, the background tissue is not suppressed and there is little, if any, contrast with the flowing blood.

There are several factors that can have an effect on flow enhancement. In multi-slice imaging, including volume acquisition, the degree of enhancement can vary with slice position. Only the first slice in the direction of flow receives fully magnetized blood. As the blood reaches the deeper slices, its magnetization and resulting signal intensity will be reduced by the RF pulses applied to the outer slices. Slowly flowing blood will be affected the most by this. Faster flowing blood can penetrate more slices before losing its magnetization. Related to this is a change in the apparent cross-sectional area of enhancement from slice to slice. A first slice might show enhancement for the entire cross section of a vessel. However, when laminar flow is present, the deeper slices will show enhancement only for the smaller area of fast flow along the central axis of the vessel. Any other effects that produce black blood can counteract flow-related enhancement. One of the most significant is the flow-void effect, which takes over at higher flow velocities.

Bright blood from flow-related enhancement is especially prevalent with gradient-echo imaging. There are two major reasons for this. The short TR values typically used in SAGE imaging increase the flow-related enhancement effect. Also, when a gradient rather than an RF pulse is used to produce the echo, there is no flow-void effect to cancel the enhancement. Therefore, gradient echo imaging is used when using the in-flow process to produce bright blood specifically, as in MR angiography.

Flow-Void Effect (Black Blood)

Relatively high flow velocities through a slice reduce signal intensity and blood brightness with spin echo imaging. Figure 12-3 illustrates this effect. The flow-void effect occurs because the blood flows out of the slice between the 90° and the 180° pulses. The arrow in the slice indicates the level of residual transverse magnetization that is present when the 180° pulse is applied to form the spin echo signal. This is the transverse magnetization produced by the preceding 90° excitation pulse. The time interval between the 90° and the 180° pulses is one-half TE. If the blood is not moving, the blood that was excited by the 90° pulse will be within the slice when the 180° pulse is applied. This results in maximum rephasing of the transverse protons at the echo event and relatively bright blood.

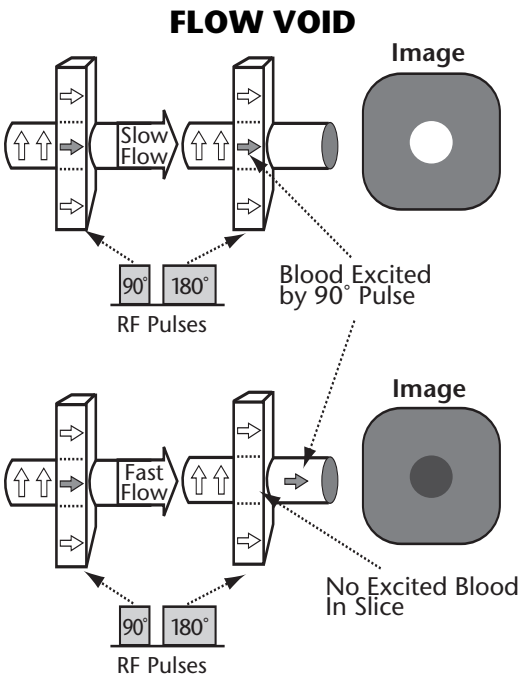


Figure 12-3. The flow-void effect caused by blood flowing out of the slice between the 90° and 180° pulses.

If the blood moves out of the slice between the 90° and 180° pulses, complete rephasing and the formation of an echo signal will not occur. This is because the 180° pulse can affect only the blood within the slice. The spin echo signal is reduced, and the flowing blood appears darker than blood moving with a lower velocity. The intensity continues to drop as flow is increased until the flow velocity removes all magnetized blood from the slice during the interval between the two pulses (one-half TE).

Selective Saturation

As we have observed on several occasions, saturation of the longitudinal magnetization with a 90° RF pulse produces a temporary darkness of a tissue or fluid because there is little magnetization to produce a signal. We saw this applied in Chapter 8 as a method for reducing artifacts from moving tissue. Selective saturation of a specific anatomical region is an effective way of producing black blood.

This technique is generally known as presaturation because the saturation pulse is applied before the imaging pulses in each cycle, and is applied to the blood before it enters the image slice. Figure 12-4 illustrates this concept. An RF pulse is selectively applied to the anatomical region that supplies blood to the slice. This pulse destroys the longitudinal magnetization by flipping it into the transverse plane. When this saturated, or unmagnetized, blood flows into the slice a short time later, it is not capable of producing a signal. Therefore, the image displays a void or black blood in the vessels. The region, or slab, of presaturation can be placed on either side of the imaged slice. This makes it possible to selectively turn off the signals from blood flowing in opposite directions.

The presaturation technique can be used to: (1) produce black-blood images; (2) selectively image either arterial or venous flow; and (3) reduce flow-related artifacts, as described in Chapter 14.

Phase Effects

There are two important phase relationships that can be affected by movement of blood during the imaging process. One is the phase relationship among the spinning protons within each individual voxel (intravoxel) and the other is the voxel-to-voxel (intervoxel) relationship of the transverse magnetizations, as shown in Figure 12-5. Both of these concepts have been described before. We will now see how they are affected by flowing blood, how they contribute to image contrast, and how they can be compensated for by the technique of flow compensation.

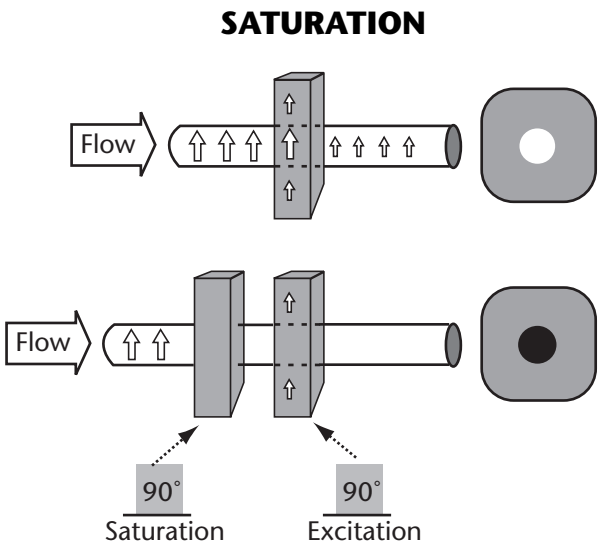


Figure 12-4. The application of a saturation pulse to eliminate the signals from flowing blood.

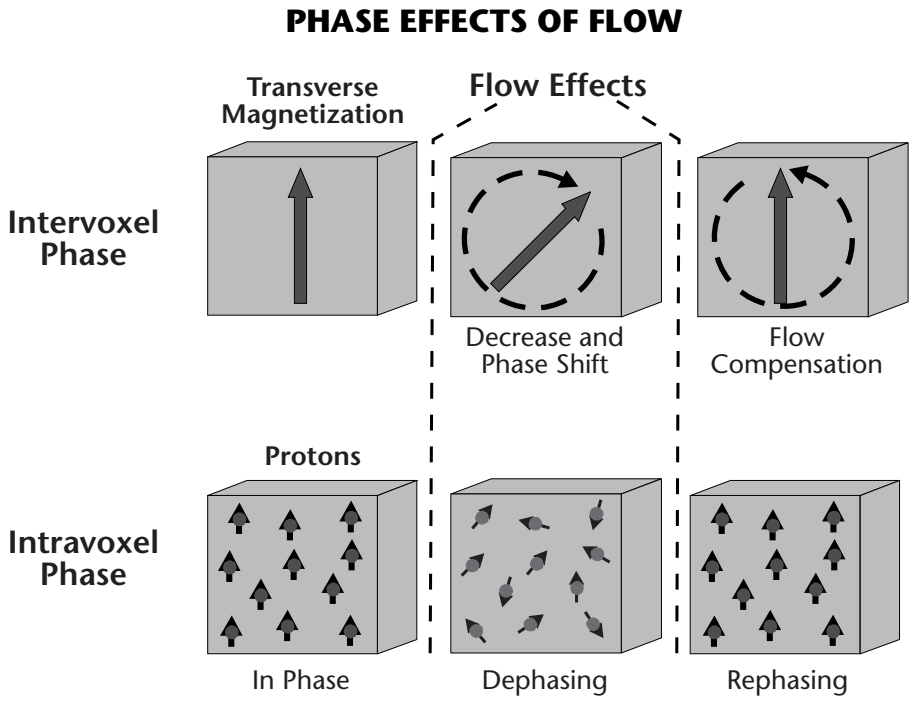


Figure 12-5. Changes in both intervoxel and intravoxel phases produced by flow.

Intravoxel Phase

In order to produce a signal, the protons within an individual voxel must be in-phase at the time of the echo event. In general, dephasing and the loss of transverse magnetization occur when the magnetic field is not perfectly homogeneous or uniform throughout a voxel. A gradient in the magnetic field is one form of inhomogeneity that produces proton dephasing. Since gradients are used for various purposes during an image acquisition cycle, this dephasing effect must be taken into account. For stationary (non-flowing) tissue or fluid the protons can be rephased by applying a gradient in the opposite direction, as shown in Figure 12-6. We saw how a gradient was used to diphase and then rephase protons to produce a gradient echo event in Chapter 7. Let us now consider this process in more detail and apply it to flowing blood.

In general, we are considering events that happen within the TE interval; that is, between the excitation pulse and the echo event. We recall that the phase-encoding gradient is applied during this time. We will use it as the example for this discussion. When the gradient is applied as shown, the right side of the voxel is in a stronger magnetic field than the left. This means that the protons on the right are spinning faster than those on the left and quickly get out of phase. However, they can be rephased by applying a gradient in the reverse direction as shown. Now the protons on the left are in the stronger field and will be spinning faster. They will catch up with and come into phase with the slower spinning protons on the right. At the time of the echo event the protons are in-phase and a signal is produced. The process just described is used in virtually all imaging methods to compensate for gradient dephasing.

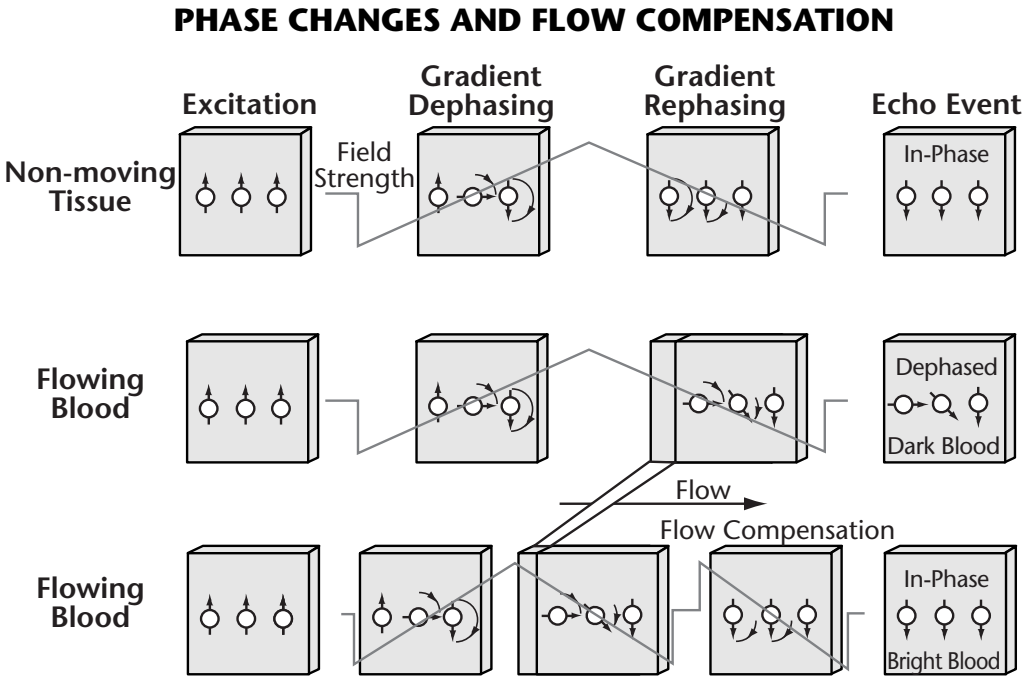


Figure 12-6. Dephasing and rephasing produced by gradients and flow compensation.

Flow Dephasing

Our next step is to consider what happens to the protons in a voxel of flowing blood or other fluid. This is also illustrated in Figure 12-6. If the voxel is moving, the protons will not be completely rephased by the second gradient. This is because the protons will not be in the same position relative to the gradient and will not receive complete compensation. The result of this is that flow in the direction of a gradient will generally produce proton dephasing and little or no signal at the time of the echo event. This is one possible source of black blood.

Flow Compensation

There are techniques that can be used to rephase the protons within a voxel of flowing blood or other fluid. These techniques use a somewhat complex sequence of gradients to produce the rephasing, as shown in Figure 12-6. The technique is generally called *flow compensation* or *gradient moment nulling*.

One of the selectable protocol factors controls the characteristics of the flow compensation gradients. The technique can be turned on or off and adjusted to compensate for specific types of flow. Flow with a constant velocity is the easiest to compensate. It is possible to compensate pulsatile (changing velocity or acceleration) flow by using a more complex gradient waveform. Flow compensation is somewhat velocity dependent. That is, all velocities are not equally compensated by the same gradient waveform.

One factor that must be considered is the time required within the TE interval for the flow compensation process. Some of the more complex compensation techniques might increase the shortest TE that can be selected. Flow compensation gradients can also be used to compensate for flow-induced *intervoxel* phase changes, which are described later.

There are two major applications of flow compensation: One is to reduce flow-related artifacts (Chapter 14) and loss of signal; and the other is as a part of phase contrast angiography, which will be developed later in this chapter.

Even-Echo Rephasing

The phenomenon of even-echo rephasing can be observed when a multi-spin echo technique is used to image blood flowing at a relatively slow and constant velocity. This effect produces an increase in signal intensity (brightness) in the even-echo images (second, fourth, etc.) compared to the odd-echo images.

This effect occurs when there is laminar flow through the vessel and the individual voxels. This means that the different layers of protons within a voxel are moving at different velocities and will experience different phase shifts as they flow through a gradient. This is a dephasing effect that will reduce transverse magnetization and signal intensity at the time of the first echo event. The key to even-echo rephasing is that the second 180° pulse reverses the direction of proton spin. Before the pulse, the protons in the faster layers had moved ahead and gained phase on the slower moving protons. However, immediately after the pulse they are flipped so that they are behind the slower-spinning protons. This sets the stage for them to catch up and come back into phase. This occurs at the time of the second echo event and results in an increase in transverse magnetization, higher signal intensity, and brighter blood than was observed in the first echo image.

The alternating of blood brightness between the odd- and even-echo images will continue for subsequent echoes, but there will be an overall decrease in intensity because of the T2 decay of the transverse magnetization. Both turbulent and pulsatile flow tend to increase proton dephasing within a voxel, which results

in a loss of signal intensity. Under some conditions this can counteract the effect of even-echo rephasing.

Intervoxel Phase

If a voxel of tissue moves during the image acquisition process, the phase relationship of its spinning transverse magnetization can be shifted relative to that of other voxels. There are both advantages and disadvantages of this effect.

Phase Imaging

Most MRI systems are capable of producing phase images. The phase imaging that is described here is different from phase contrast angiography that will be described later. A phase image is produced with one of the conventional imaging methods but with a different way of calculating the image from the signals. A phase image is one in which the brightness of a voxel is determined by its phase relationship rather than the magnitude of transverse magnetization. This is illustrated in Figure 12-7.

Consider a row of voxels across a vessel through which blood is flowing. We will assume laminar flow with the highest velocity along the central axis of the vessel. As this blood flows through a magnetic field gradient, the phase of the individual voxels will shift in proportion to the flow velocity. There, if we create an image at a specific time, we can observe the phase relationships. If the phase of a voxel's transverse magnetization is then translated into image brightness (or perhaps color), we will have an image that displays flow velocity and direction. This type of image is somewhat analogous to a Doppler ultrasound image in that it is the velocity that is being measured and displayed in the image.

Analytical software can be used in conjunction with phase images to calculate selected flow parameters.

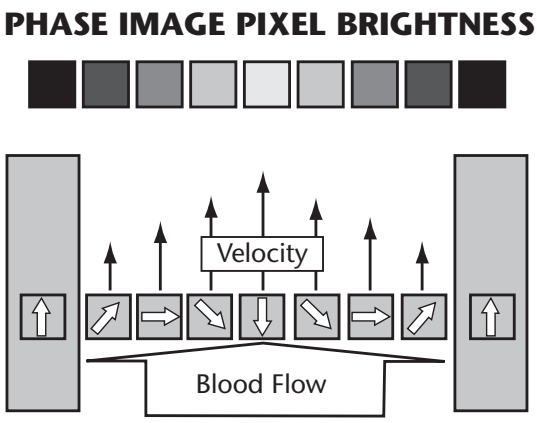


Figure 12-7. Intervoxel phase changes can be used to measure and image blood flow.

Artifacts

Flow and other forms of motion can produce serious artifacts in MR images. Although the techniques that can be used to suppress artifacts will be described in Chapter 14, it is appropriate to consider the source of one such artifact here. We recall that the process of phase-encoding is used to produce one dimension in the MR image. A gradient is used to give each voxel in the phase-encoded direction a different phase value. This phase value of each voxel is measured by the reconstruction process (Fourier transform) and used to direct the signals to the appropriate image pixels, as described in Chapter 9. This process works quite well if the tissue voxels are not moving during the acquisition process. However, if a voxel moves through a gradient, the phase relationship of its transverse magnetization will be altered, as described above. This means that the RF signal will no longer carry the correct phase address and will be directed to and displayed in the wrong pixel location. The observable effect is streaking or ghost images in the phase-encoded direction. Although any type of tissue motion can produce this type of artifact, it is a

significant problem with flowing blood. Flow compensation as described above, and other techniques to be described in Chapter 14 are used to reduce this type of artifact.

Angiography

The MR angiography methods use a combination of the effects described above to produce vascular images. Generally, MR angiography is used to produce vascular images covering a thick anatomical volume as opposed to a relatively thin slice as in most other imaging methods. There are several different approaches used to produce MR angiograms. Each method has specific characteristics that must be considered when producing clinical images. The three general methods based on how the contrast between blood and background tissue are produced are:

1. Phase Contrast Angiography.
2. In-Flow Contrast Angiography.
3. Contrast Enhanced Angiography.

The first two methods derive contrast from the movement of the blood and the flow effects that have just been described. The third method uses administered contrast media to enhance the vascular contrast.

Phase Contrast Angiography

We recall that when blood flows through a magnetic-field gradient, the phase relationship is affected. This applies both to the phase relationship of protons within a voxel and the voxel-to-voxel relationship of transverse magnetization. Both of these—intravoxel and intervoxel—phase effects can be used to produce contrast of flowing blood. However, the intervoxel phase shift of the transverse magnetization is the effect most frequently used to produce phase contrast angiograms.

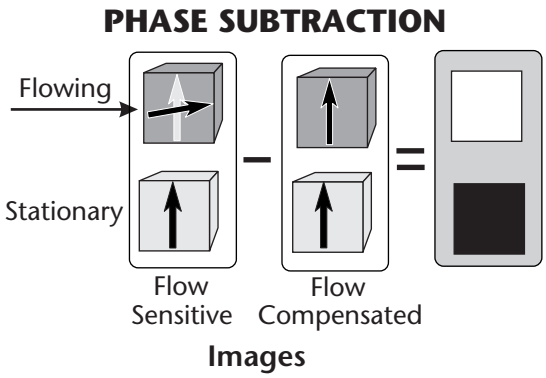


Figure 12-8. The principle of phase contrast angiography. Blood brightness is related to the phase change produced by flow.

When blood flows through a gradient, the phase of the transverse magnetization changes in proportion to the velocity. When this technique is used, the phase shift, not the magnitude of the magnetization, is used to create the image. Therefore, blood brightness is directly related to flow velocity in the direction of the flow-encoding gradient.

Figure 12-8 illustrates the basic process of creating a phase contrast angiogram. At least two image acquisitions are required. One image is acquired with a flow phase-encoding gradient turned on. The phase of the magnetization is shifted in proportion to the flow velocity. A flow compensation gradient is applied during the acquisition of a second image to reset the phase of the flowing blood. The phase in the stationary tissue is not affected and is the same in both images.

The mathematical process of phase or vectory subtraction used to produce the phase contrast image is illustrated in Figure 12-9. The vector subtraction process determines the difference in phase between the flow-encoded blood and the flow-compensated blood, which serves as a reference. The phase difference (and

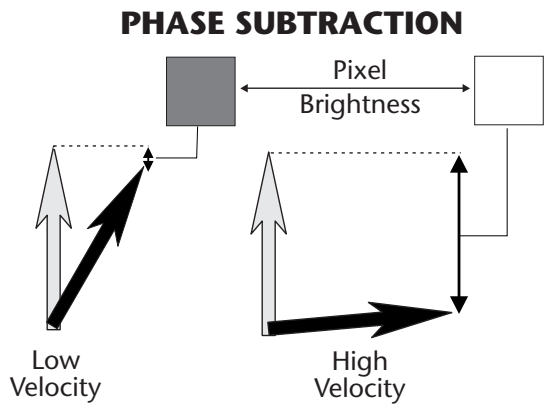


Figure 12-9. The principle of phase subtraction.

blood brightness) increases with velocity. However, when the phase reaches the 6 o'clock position, additional increases in velocity will produce a reduced difference in phase and brightness as the phase returns to the 12 o'clock reference position. The contrast in this type of image is related to both the velocity and direction of the flowing blood.

Velocity

Because the degree of phase shift and the resulting contrast is related to flow velocity in a cycling manner, the operator must set a velocity value as one of the protocol factors. This will give a specific relationship between phase values and velocity for that particular imaging procedure. Flow at this rate will produce maximum contrast as shown in Figure 12-10. The important thing to note is that at some velocities, both below and above the set velocity, the blood will be black. This can produce an aliasing artifact in which fast-flowing blood will be dark and will look like, or alias, slow-flowing blood. This must be taken into account both when selecting a set velocity value and when viewing images. An advantage of phase contrast angiography is the ability to image specific ranges of flow, including relatively slow flow rates if the proper factors are used.

Flow Direction

Phase contrast is produced only when the flow is in the direction of a gradient. To image

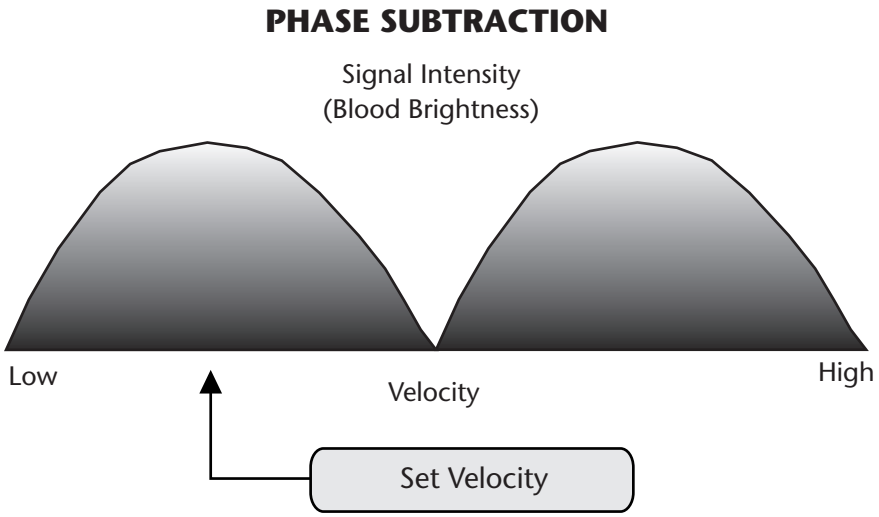


Figure 12-10. The relationship of blood brightness to flow velocity in phase contrast angiography.

blood that is flowing in different directions, several image data sets must be acquired with gradients in the appropriate directions. Flow in all possible directions can be imaged by acquiring three images with three orthogonal gradient directions.

The images for the different flow directions are combined with the subtraction process to produce one composite angiographic image.

In-Flow Contrast Angiography

With the in-flow contrast method an angiographic image is created by using the flow-related enhancement principle described previously. This is also called the time-of-flight technique. Bright blood images are produced by using a gradient echo method with relatively short TR values. Because of the short TR values, the background tissue remains partially saturated (and dark) while the flowing blood produces a stronger (bright) signal because of the in-flow effect.

To produce an angiogram, the method must be extended to cover an anatomical volume rather than a single slice. There are two acquisition techniques that can be used to achieve this: 3-D volume acquisition or multiple 2-D slice acquisitions. Each has its special characteristics that must be considered in clinical applications.

Three-Dimension (3-D) Volume Acquisition

A 3-D volume acquisition can be used to create an angiographic image, as shown in Figure 12-11. We recall (from Chapter 9) that with this acquisition method RF pulses are applied to and signals are acquired from an entire volume simultaneously. During acquisition a phase-encoding gradient is applied in the slice selection direction. The individual slices are then created during the image reconstruction process. The principal advantage is the ability to create thin, contiguous slices with small voxel dimensions. This gives good image

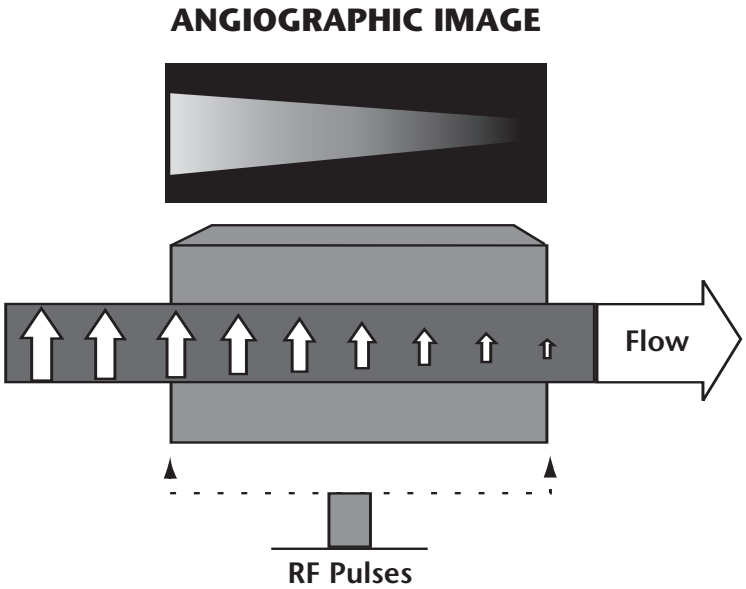


Figure 12-11. A 3-D volume acquisition showing decreased magnetization and blood brightness produced by saturation as the blood flows deeper into the volume.

detail. Also, in vascular imaging small voxels reduce intravoxel dephasing and signal loss.

The principal disadvantage of this method is that relatively slow-flowing blood becomes saturated as it passes through the acquisition volume. This can limit the volume size and vessel length, which is capable of producing good image contrast. Blood flowing with a relatively high velocity, as along the central axis of a vessel, will be imaged deeper into the acquisition volume than slow-flowing blood. This is illustrated in Figure 12-11.

Two-Dimension (2-D) Slice Acquisition

With this technique the anatomical volume is covered by acquiring a series of single-slice images, as shown in Figure 12-12. This approach provides relatively strong signals (bright blood) throughout the volume because each slice is imaged independently and not affected by blood saturated in other slices. This makes it possible to image relatively long vessels extending over large volumes. Also, with this method, the signals from the flowing blood are relatively independent of flow velocity.

Image Format

The image acquisition and reconstruction process results in data in the form of many slice images covering a volume within the patient's body. This must then be reformatted into a 3-D image of the vascular structure extending over this volume. There are several methods that can be used for this purpose.

Maximum Intensity Projection (MIP)

The maximum intensity projection (MIP) technique is commonly used to create a single com-

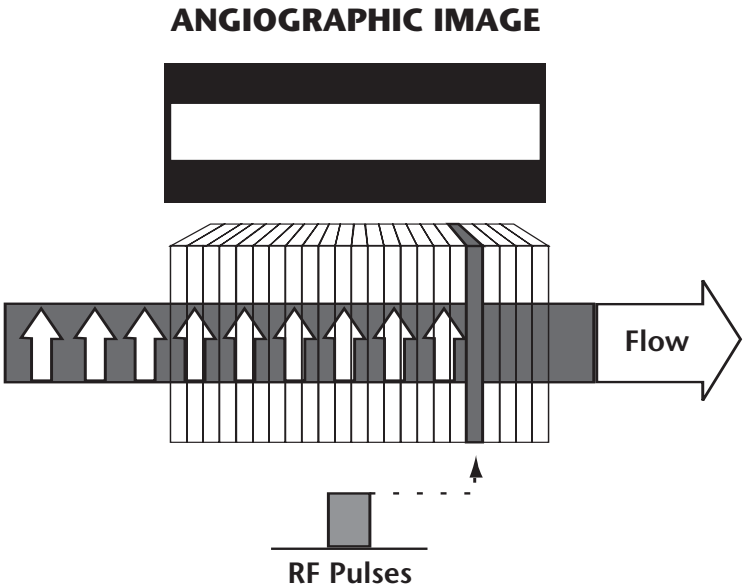


Figure 12-12. The acquisition of thin slices reduces the saturation and loss of signal intensity within the imaged volume. Compare to figure 12-11.

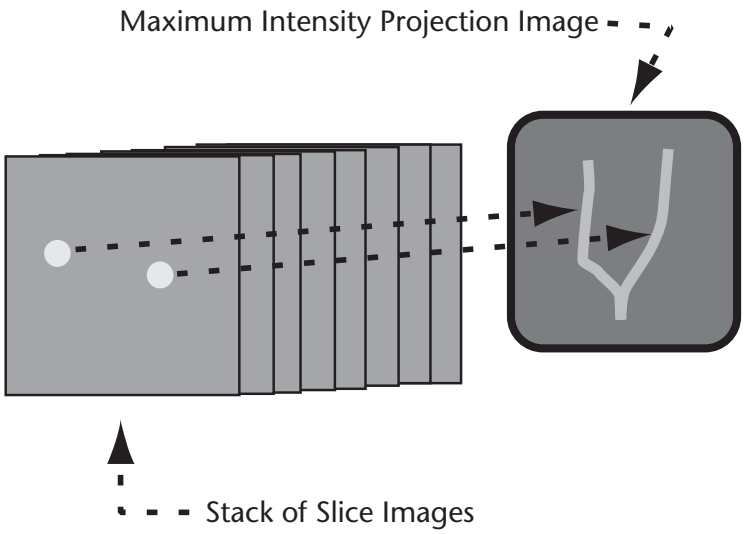


Figure 12-13. The use of maximum intensity projection (MIP) to produce a composite image from a stack of slice images.

posite image from a stack of images, as shown in Figure 12-13. This is a mathematical process performed by the computer. The stack of images is "viewed" along a series of pathways through the volume. The maximum signal intensity, or blood brightness, encountered in any slice along each pathway is then projected onto the composite image. The resulting MIP image is a 2-D image of the 3-D vascular structure. This process generally enhances the contrast between the flowing blood and the stationary tissues.

It is possible to create images by projecting in different directions through the volume. This gives the impression of rotating the vascular structure so that it can be viewed from different directions.

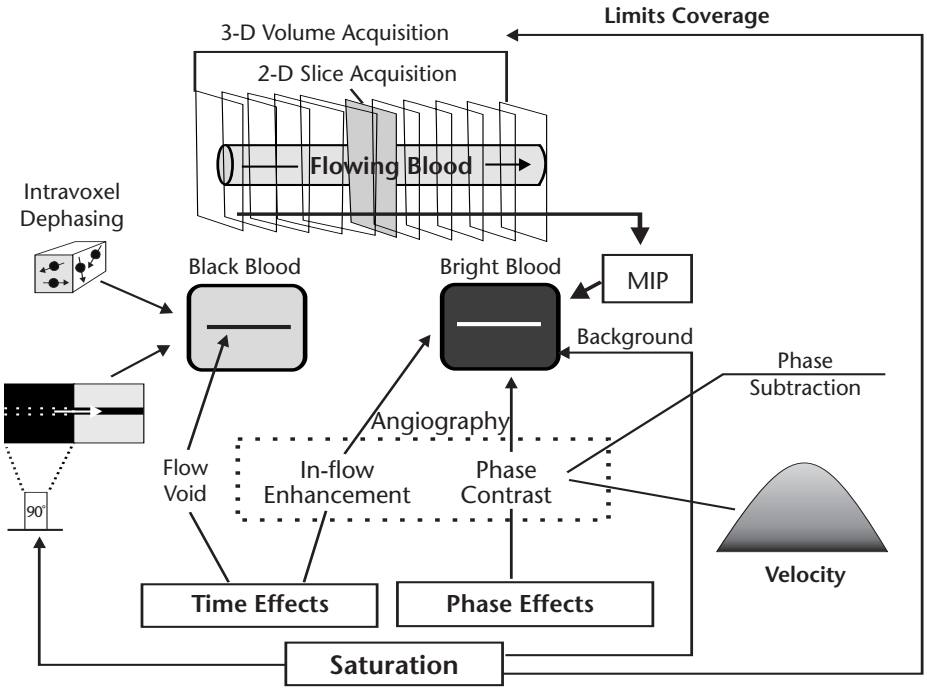
Surface Rendering

Images can also be formed by using surface rendering programs. This is especially useful for large vessels such as the aorta.

Contrast Enhanced Angiography

Images of vascular structures can be produced by injecting contrast media and not relying on the time and phase effects described previously to produce the image contrast. This overcomes some of the problems of signal loss produced by saturation in large areas and the limited velocity range than can be imaged with phase contrast. It is, however, restricted in terms of depending on the presence of contrast media in the vessels for continuous imaging.

Mind Map Summary
Vascular Imaging



MR is capable of producing images of flowing blood without the injection of contrast media. The visibility or contrast of the blood is produced by a variety of physical effects or interactions between the moving blood and the MR process. Some effects produce black blood and others produce bright blood. Black blood occurs when the flowing blood does not produce a signal. This can be caused by the flow void effect, the application of saturation pulses to a vascular area, or any condition that produces intravoxel dephasing.

Bright blood effects, used in angiography, are in-flow enhancement and phase contrast. In-flow enhancement produces bright blood by keeping the surrounding stationary tissue partially saturated. Blood flowing into the image area is not saturated and produces a bright signal. However, consideration must be given to how the image slices are formed and acquired over an anatomical region. 3-D volume acquisition produces thin slices and high image detail, but results in partial saturation of the flowing blood and possible deterioration of the image. An alternative method that produces less saturation of the flowing blood is the 2-D slice acquisition process.

In phase contrast angiography the transverse magnetization of flowing blood experiences a phase shift when passing through a gradient. This phase change, which is proportional to velocity, can be measured and translated into brightness in an image. Most MR angiographic images represent a 3-D anatomical region and are produced from a stack of slice images by the MIP process or by surface or volume rendering image processing.

13

Functional Imaging

Introduction And Overview

MR functional imaging consists of several different imaging methods that are used to visualize and, in some cases, quantify blood and fluid movement beyond the general vascular system. In the last chapter we learned that there are several techniques that can be used to produce contrast and a visible image of blood flowing through vessels. It is the perfusion throughout the capillary bed and then the diffusion of fluids throughout the tissue that is the subject of most MR functional imaging procedures. Specific and different methods are used to image both the perfusion and diffusion processes and to visualize tissue areas that are metabolically active.

Diffusion Imaging

Diffusion imaging is generally used to visualize tissue areas in which some pathologic process has altered the motion of fluid that is outside of the vessels and capillaries. This is the natural, random, incoherent, Brownian motion of the proton-containing molecules. The motion is always there, but its rate is what changes and is the source of contrast in diffusion imaging.

Brownian Motion and Diffusion

Free or unbound molecules in a material such as fluid are in constant motion because of thermal activity. This is a random motion in which the molecules travel in one direction, collide with other molecules and change direction,

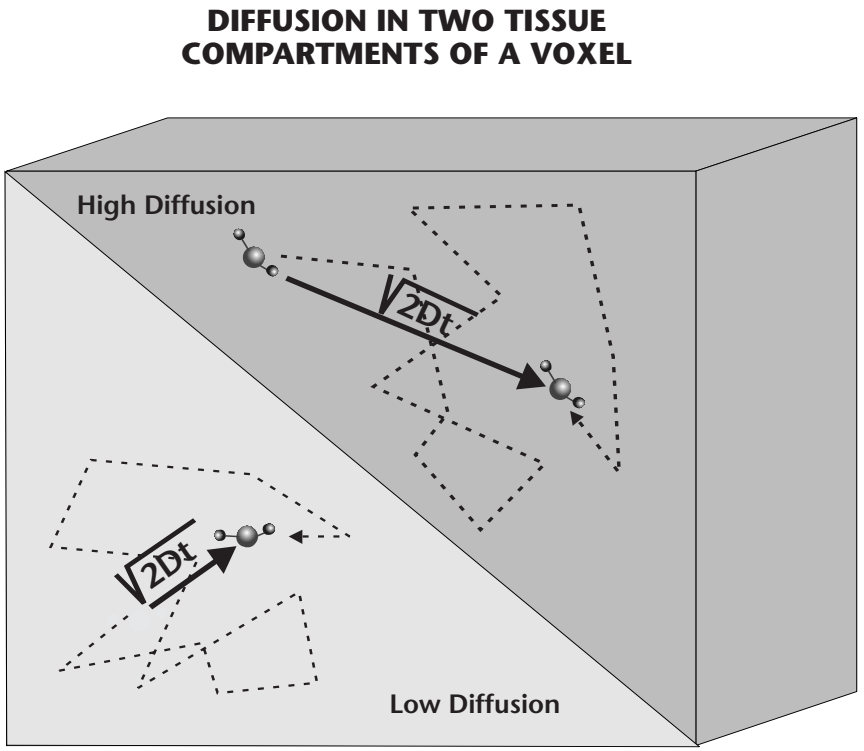


Figure 13.1. The average distance traveled by a molecule in time, t, depends on the diffusion coefficient (D) for the specific tissue compartment within a voxel.

and then move in another direction until the next collision. This is known as Brownian motion and is illustrated in Figure 13-1. It is this motion that is the source of the general process of diffusion. Diffusion is what accounts for the net movement of a substance from an area of high concentration to an area of lower concentration in the absence of other pressures or forces. Within a region with a uniform concentration, such as a voxel, the molecules are still in motion even if there is no net movement from one region to another.

The distance that the molecules move in a specific time is very dependent on structural characteristics of this tissue. The motion is generally greatest in fluids and somewhat less in a cellular tissue environment.

Diffusion Coefficient

The diffusion coefficient, D, is related to the net displacement of molecules in a given time. The net displacement distance is much less than the total path length traveled by a molecule during this time. This is a statical process in which there is a range of distances traveled by molecules in the same period of time. The mean (rms) distance traveled is what is related to the diffusion coefficient as shown in Figure 13-1.

Apparent Diffusion Coefficient (ADC)

Diffusion coefficient values determined by MRI might be a composite from several structural

compartments (extracellular and intracellular) within a voxel. There could be different diffusion coefficient values within these various compartments. Therefore, the values determined are designated as the apparent diffusion coefficient, ADC.

Diffusion Direction

In a specific tissue the diffusion rate might be different in different directions because of the orientation of certain tissue structures. This is an important factor that must be taken into account when producing diffusion images. As we will soon discover, the sensitivity for observing diffusion is produced by applying magnetic field gradients. Only when a gradient is applied in the direction of the diffusion will it be imaged.

The Imaging Process

Diffusion imaging is based on the principle that the diffusion motion of the molecules produces a dephasing of the spinning protons within a voxel and that this results in a reduced signal intensity and image brightness. The dephasing is actually produced by applying additional diffusion-sensitizing gradients during the image acquisition cycle as shown in Figure 13-2. Here we see the gradients used in conjunction with a spin echo pulse sequence. Notice that there are actually two gradient pulses. One is applied before the 180° RF pulse and the second gradient is applied after the pulse. Let us recall that a gradient is a variation in field strength across each individual voxel. During the time of the gradient the spinning protons will be in different field strengths and

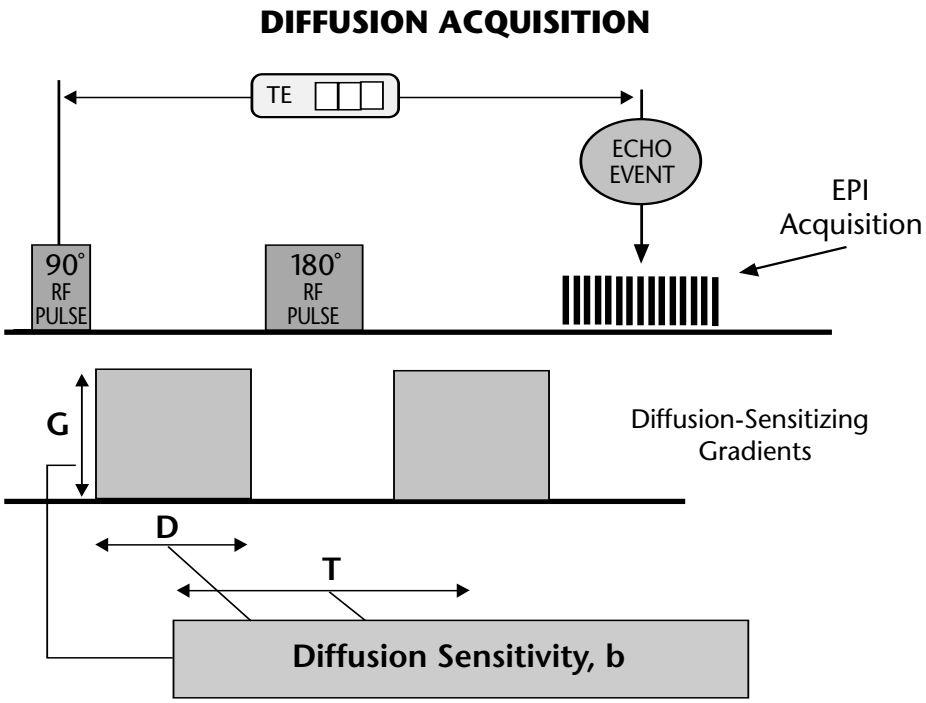


Figure 13.2. The application of gradients to produce dephasing and reduced signal intensity related to diffusion.

spinning at different rates, along the direction of the gradient. This produces a dephasing of the protons within the voxel. When the 180° RF pulse is applied, it reverses the spin direction. Now when the second gradient is applied, it produces a rephasing on the spinning protons within the voxel. However, only the protons that have not moved or changed position between the times of the two gradients will be completely rephased. The protons in molecules that have moved will be in a different location and field strength during the second gradient and will not be completely rephased. This results in reduced signal intensity from voxels containing the moving protons. It is this reduction in signal intensity that produces the contrast of the diffusing molecules with respect to the non-moving tissue structures.

The reduction in signal intensity (S_D/S_0) produced by the diffusion depends on two factors: the rate of diffusion expressed by the value of the diffusion coefficient, D, and the diffusion sensitivity, b, which is determined by characteristics of the gradients. The relationship is given by:

$$S_D/S_0 = e^{-bD}$$

Diffusion Sensitivity

The diffusion sensitivity, b, is determined by the strength, duration, and time separating the

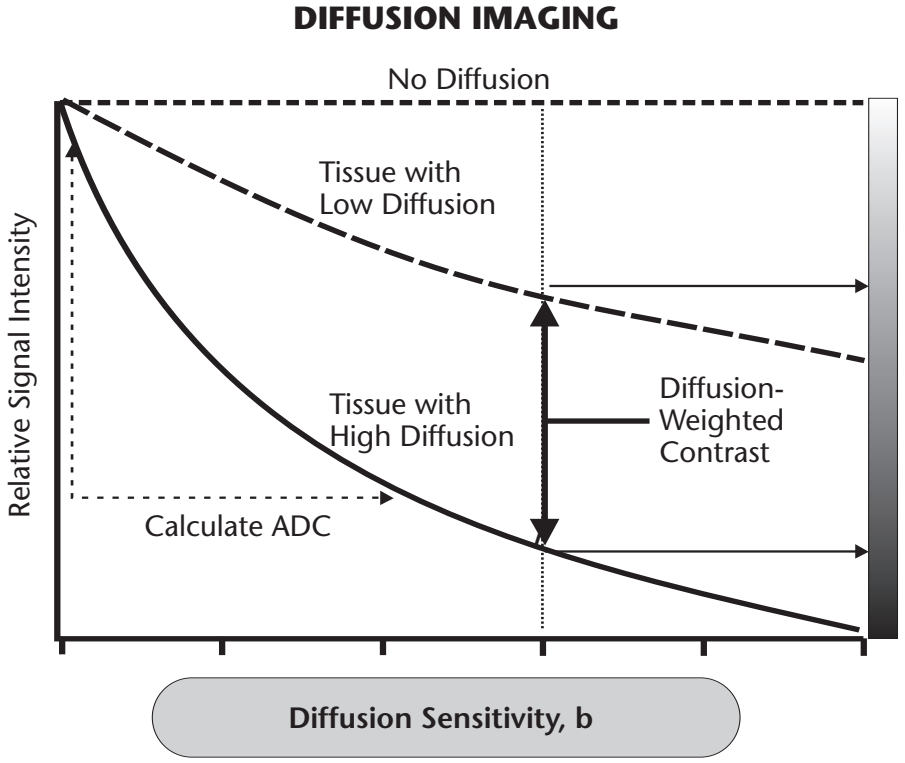


Figure 13-3. The relationship of signal intensity to the diffusion rates in tissues and the setting of the diffusion sensitivity protocol parameter.

two gradients. These factors are identified in Figure 13-2. The sensitivity is very dependent on the gradient strength, G. Maximum gradient strength is a design characteristic of the imaging system. Diffusion imaging is generally limited to systems equipped with strong, high-performance gradients.

The value of the diffusion sensitivity, b, is an adjustable protocol factor. Images can be acquired with different sensitivity values.

Diffusion-Weighted Images

Images in which the diffusion rate is a source of contrast can be obtained by applying the diffusion-sensitizing gradients as described. The sensitivity can be adjusted by changing the b protocol factor. As the value of b is increased, diffusion has an increased weighting and areas with relatively high levels of diffusion become darker. This is illustrated in Figure 13-3.

ADC Map Images

An alternative to diffusion-weighted images is to calculate the apparent diffusion coefficient, ADC, for each voxel and display the values in the form of an image. This can be done because the slope of the signal intensity versus b value curve is determined by the ADC of each specific tissue voxel. By acquiring images at several (at least two) b values, the ADC can be calculated. ADC map images have contrast that is opposite to diffusion-weighted images. Areas of increased diffusion are bright.

Blood Oxygenation Level Dependent (BOLD) Contrast

Blood oxygenation level dependent, or BOLD, is a source of contrast related to neural activity that is endogenous and does not require the administration of any contrast agent. Advantages are

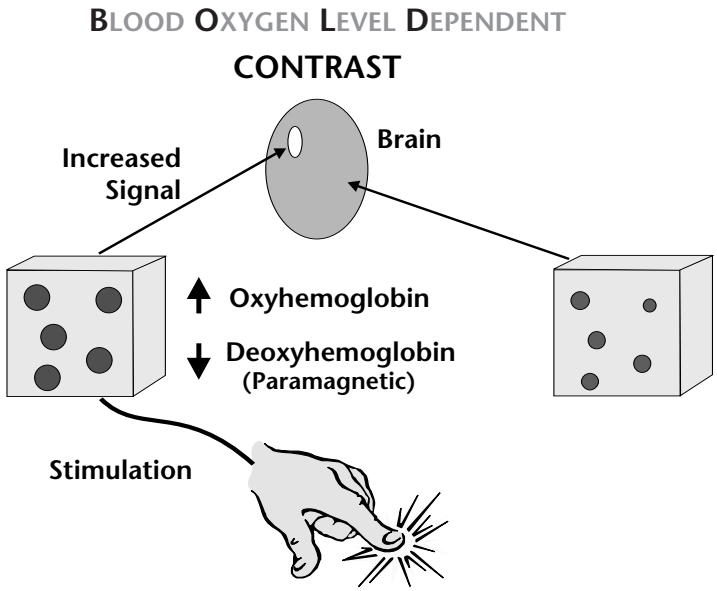


Figure 13-4. The BOLD effect for visualizing areas of the brain activated by stimulation.

that it is not invasive and can be used continuously to observe dynamic effects.

BOLD is based on the fact that deoxyhemoglobin is paramagnetic but oxyhemoglobin is not. This means that deoxyhemoglobin has a higher magnetic susceptibility and produces more local field inhomogeneities than oxyhemoglobin does. This results in more rapid proton dephasing in areas of high deoxyhemoglobin concentration.

The BOLD technique is used to visualize activated areas in the brain through a series of events as illustrated in Figure 13-4. Brain activity produces an increase in metabolism and oxygen consumption. The associated vasodilatation results in an increase in blood flow. When the flow (oxygen delivery) increases more than the oxygen consumption, there is an increase in the local oxygen level. The increase in oxyhemoglobin in relationship to the deoxyhemoglobin reduces the susceptibility effect and rate of proton dephasing. This results in an increase in the local T2* value.

When imaged with a T2*-weighted acquisition, activated brain areas will produce an increased signal intensity and brightness. The increase in signal intensity is small, on the order of a few percent. However, by taking the difference between images acquired *with* and *without* activation, the areas of activation can be observed.

Perfusion Imaging

Blood flow through the microvasculature, or tissue perfusion, can be evaluated with MRI.

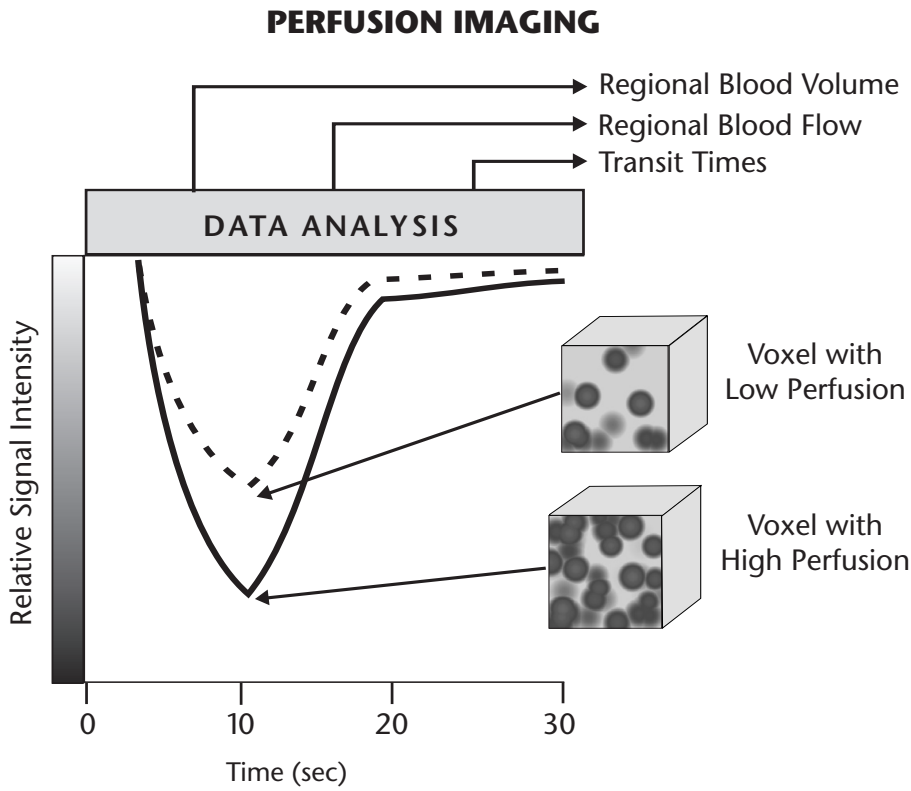


Figure 13-5. The reduction in signal intensity produced by the passage of a bolus of contrast media provides information on tissue perfusion.

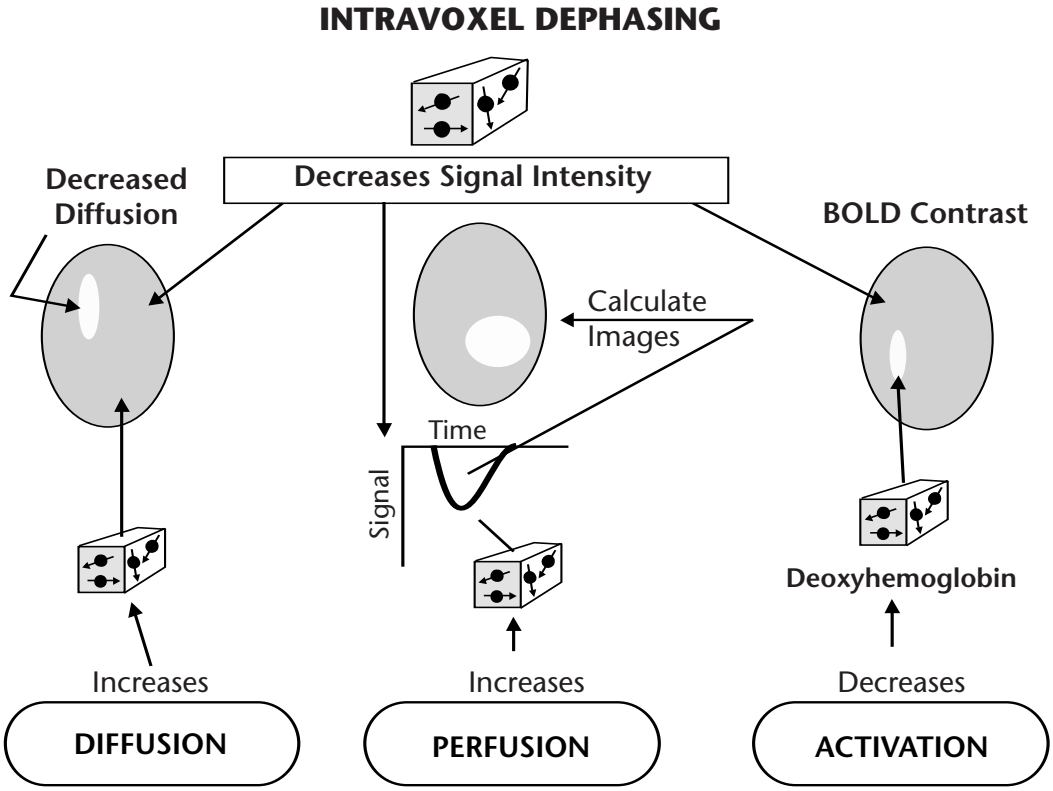
In the previous chapter we learned that when blood is flowing through relatively large vessels, compared to capillaries, there are several physical effects (time effects and phase changes) that can be used as sources of contrast and vessel visualization. These effects do not work for the small capillaries that distribute the blood throughout the tissues. Therefore, other methods for producing contrast must be used. The objective is not to visualize the flow in individual vessels, as in angiography, but to evaluate the combined flow through many microvessels within a volume of tissue such as a voxel.

A common method for imaging perfusion is illustrated in Figure 13-5. The process is started by administering a bolus of contrast media to the patient. As the bolus passes through a specific tissue, there will be an increase in local magnetic susceptibility. This produces a decreased signal intensity in T2*-weighted images. Voxels with the highest perfusion will experience the largest reduction in signal intensity as the bolus passes through.

The reduction in signal intensity with the passage of the bolus through two different tissues is illustrated in Figure 13-5. Acquisition of perfusion data requires fast imaging methods, such as EPI, because images must be acquired every few seconds to properly measure the characteristics of the bolus passage.

The area under the signal versus time curve is proportional to the regional blood volume. By measuring the transit time of the bolus information on the regional blood, flow can be obtained.

Mind Map Summary
Functional Imaging



Diffusion imaging is based on the principle that the diffusion motion of molecules produces intravoxel dephasing when appropriate gradients are applied. The diffusion sensitizing gradients can be adjusted to produce different levels of diffusion sensitivity, making it possible to measure the apparent diffusion coefficient values in voxels of tissue. These values can be displayed as a map image. A diffusion-weighted image can also be produced in which areas with decreased diffusion will appear bright.

Perfusion imaging is achieved by injecting a bolus of contrast media. As the bolus passes through the imaged tissue, the intravoxel dephasing is increased and the signal intensity drops. When the data is collected as a rapid sequence of dynamic images, several different perfusion parameters can be calculated and displayed as images.

BOLD is an imaging method that can be used to visualize activated areas within the brain. The contrast is produced by a shift in the deoxyhemoglobin-oxyhemoglobin ratio resulting from the vasodilatation in the activated area. The signal intensity in the activated area, with increased oxyhemoglobin, increases because deoxyhemoglobin is a paramagnetic substance that contributes to intravoxel dephasing.

14

Image Artifacts

Introduction And Overview

There are a variety of artifacts that can appear in MR images. There are many different causes for artifacts, including equipment malfunctions and environmental factors. However, most artifacts occur under normal imaging conditions and are caused by the sensitivity of the imaging process to certain tissue characteristics such as motion and variations in composition.

There are many techniques that can be applied during the image acquisition process to suppress artifacts. In this chapter we will consider the most significant artifacts that degrade MR images and how the various artifact suppression techniques can be employed.

An artifact is something that appears in an image and is not a true representation of an object or structure within the body. Most MRI artifacts are caused by errors in the spatial encoding of RF signals from the tissue voxels. This causes the signal from a specific voxel to be displayed in the wrong pixel location. This can occur in both the phase-encoding and frequency-encoding directions, as shown in Figure 14-1. Errors in the phase-encoding direction are more common and larger, resulting in bright streaks or ghost images of some anatomical structures. Motion is the most common cause, but the aliasing effect can produce ghost images that fold over or wrap around into the image.

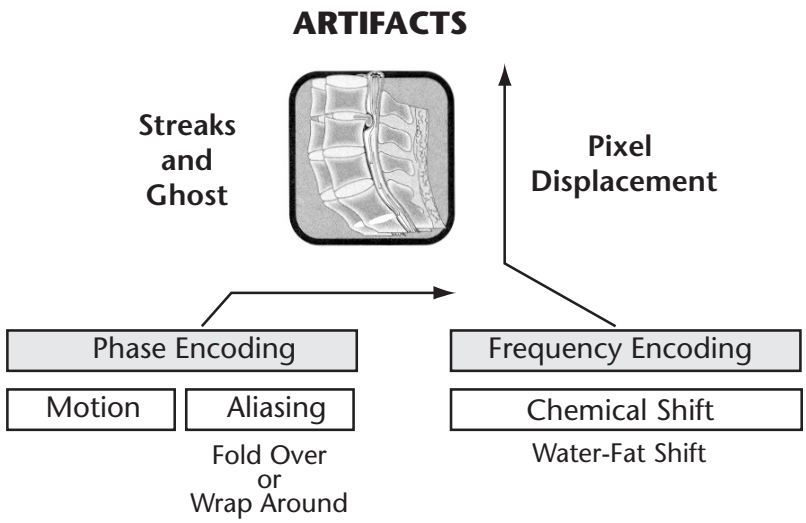


Figure 14-1. Classification of the most common MRI artifacts.

Errors in the frequency-encoding direction are limited to a displacement of just a few pixels that can occur at boundaries between fat and nonfat tissues in which most of the protons are contained in water.

Motion-Induced Artifacts

Movement of body tissues and the flow of fluid during the image acquisition process is the most significant source of artifacts. The selection of a technique that can be used to suppress motion artifacts depends on the temporal characteristic of the motion (periodic or random) and the spatial relationship of the moving tissue to the image area. Figure 14-2 shows the types of motion that can produce artifacts and techniques that can be used to reduce them.

At this point we need to make a clear distinction between blurring and artifacts. We recall that the principal source of blurring in MRI is the size of the individual voxels. Under some imaging conditions motion can produce additional blurring, which reduces visibility of detail and gives the image an unsharp appearance. This is especially true for motion that causes a voxel to change location from one acquisition cycle to another. Blurring occurs when the signals from an individual voxel come from the different locations occupied by the voxel. The signals are smeared over the region of movement. Artifacts, or ghost images, occur when the signals are displayed at locations that were never occupied by the tissue.

Most of the motion-induced artifacts are produced by dephasing or phase errors. We recall from Chapter 12 that flow and other forms of motion can produce both intravoxel and intervoxel phase problems. Intravoxel dephasing generally results in reduced signal intensity, whereas intervoxel phase errors produce artifacts in the phase-encoded direction. The phase-encoding process is especially affected by body motion, which causes a particular anatomical structure to be in a different location from one acquisition cycle to another. This contributes to the phase error and to the production of artifacts.

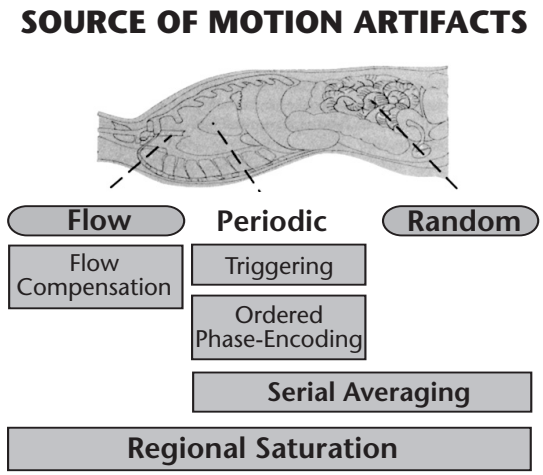


Figure 14-2. Various types of motion that produce artifacts in MR images and correction techniques.

Phase-Encoded Direction

Motion artifact streaks and ghosting always occur in the phase-encoded direction. Prior to an acquisition the operator can select which direction in the image is to be phase-encoded as opposed to frequency-encoded. This makes it possible to place the artifact streaks in specific directions. This is a very helpful technique for protecting one anatomical area from motion and flow artifacts produced in another area. It does not eliminate the artifacts but orients them in a specific direction.

Cardiac Motion

Cardiac activity is a source of motion in several anatomical locations. The movement of the heart produces artifact streaks through the thoracic region and blurring of cardiac structures when the heart is being imaged. In other parts of the body, pulsation of both blood and cerebrospinal fluid (CSF) can produce artifacts and loss of signal intensity.

Triggering

Synchronizing the image acquisition cycle with the cardiac cycle is an effective technique for reducing cardiac motion artifacts. An EKG (electrocardiogram) monitor attached to the patient provides a signal to trigger the acquisition cycle.

The R wave is generally used as the reference point. The initiation of each acquisition cycle is triggered by the R wave. Therefore, an entire image is created at one specific point in the cardiac cycle. This has two advantages. The motion artifacts are reduced and an unblurred image of the heart can be obtained. The delay time between the R wave and the acquisition cycle can be adjusted to produce images throughout the cardiac cycle. This is typically done in cardiac imaging procedures.

Maximum artifact suppression by this technique requires a constant heart rate. Arrhythmias and normal heart-rate variations reduce the effectiveness of this technique.

Cardiac triggering is also useful for reducing artifacts from CSF pulsation. This can be especially helpful in thoracic and cervical spine imaging.

Flow Compensation

The technique of flow compensation or gradient moment nulling was described in Chapter 12. In addition to compensating for blood flow effects, this technique can be used to reduce problems arising from CSF pulsation, especially in the cervical spine. It actually provides two desirable effects. The rephasing of the protons within each voxel increases signal intensity from the CSF, especially in T2-weighted images. It also reduces the motion artifacts.

Respiratory Motion

Respiratory motion can produce artifacts and blurring in both the thoracic and abdominal

regions. Several techniques can be used to suppress these motion effects.

Averaging

The technique of signal averaging is used primarily to reduce signal noise, as described in Chapter 10. However, averaging has the additional benefit of reducing streak artifacts arising from motion. If a tissue voxel is moving at different velocities and in different locations during each acquisition cycle, the phase errors will be different and somewhat randomly distributed. Averaging the signals over several acquisition cycles produces some degree of cancellation of the phase errors and the artifacts. There are several different ways that signals can be averaged. Serial rather than parallel averaging gives the best artifact suppression. Serial averaging is performed by repeating two or more complete acquisitions and averaging. Parallel averaging is performed by repeating an imaging cycle two or more times for each phase-encoded gradient step. With serial averaging there is a much longer time between the measurements made at each phase-encoded step. This gives a more random distribution of phase errors and better cancellation. As with noise, increasing the number of signals averaged (NSA) reduces the intensity of artifacts but at the cost of extending the acquisition time. The averaging process reduces artifacts but not motion blurring.

Ordered Phase-Encoding

An artifact reduction technique that is used specifically to compensate for respiratory motion is ordered phase-encoding. We recall that a complete acquisition requires a large number of phase-encoded steps. The strength of the phase-encoded gradient is methodically changed from one step to another. In a normal acquisition the gradient is turned on with maximum strength during the first step and is gradually decreased to a value of zero at the midpoint of the acquisition process. During the second half the gradient strength is increased, step by step, but in the opposite direction. The basic problem is that two adjacent acquisition cycles might catch a voxel of moving tissue in two widely separated locations. The location is also somewhat random from cycle to cycle. This contributes to the severity of the artifacts.

Ordered phase-encoding is a technique in which the strength of the gradient for each phase-encoded step is related to the amount of tissue displacement at that particular instant. This requires a transducer on the patient's body to monitor respiration. The signals from the transducer are processed by the computer and used to select a specific level for the phase-encoded gradient.

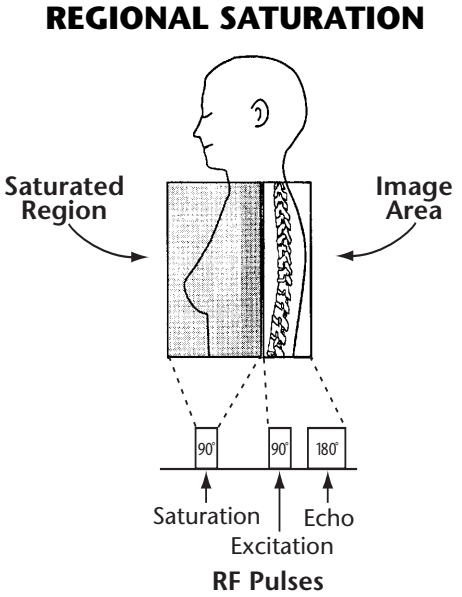


Figure 14-3. The use of regional presaturation to reduce motion artifacts.

Regional Presaturation

Regional presaturation is a technique that has several different applications. In Chapter 12 we saw how it could be used with flowing blood to eliminate the signal and produce a black-blood image. An application of this technique to reduce respiratory and cardiac motion artifacts in spine imaging is shown in Figure 14-3. With this technique a 90˚ RF saturation pulse is selectively applied to the region of moving tissue. This saturates or reduces any existing longitudinal magnetization to zero. This is then followed by the normal excitation pulse. However, the region that had just experienced the saturation pulse is still demagnetized (or saturated) and cannot produce a signal. This region will appear as a black void in the image. It is also incapable of sending artifact streaks into adjacent areas.

Flow

Flow is different from the types of motion described above because a specific structure does not appear to move from cycle to cycle. This reduces the blurring effect, but artifacts remain a problem.

The flow of blood or CSF in any part of the body can produce artifacts because of the phase-encoding errors. Several of the techniques that have already been described can be used to reduce flow-related artifacts.

Regional Presaturation

Regional presaturation, as described in Chapter 12, is especially effective because it turns the blood black. Black blood, which produces no signal, cannot produce artifacts.

Figure 14-4 illustrates the use of presaturation to reduce flow artifacts. The area of saturation is located so that blood flows from it into the image slice.

Flow Compensation

Flow compensation is useful when it is desirable to produce a bright-blood image. It both reduces intervoxel phase errors, the source of

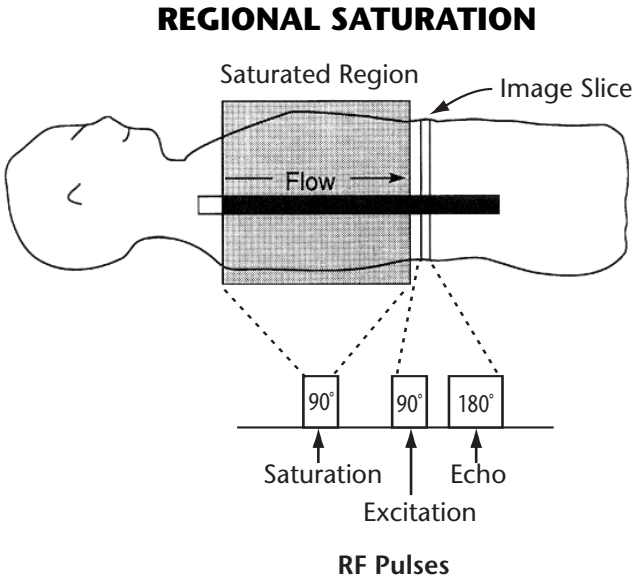


Figure 14-4. The use of regional saturation to reduce flow artifacts.

streaking, and restores some of the intravoxel magnetization and signal intensity.

Aliasing Artifacts

Aliasing, which produces foldover or wraparound artifacts, can occur when some part of the patient's body extends beyond the selected field of view (FOV). The anatomical structures that are outside of the FOV appear to "wrap around" and are displayed on the other side of the image, as shown in Figure 14-5. This occurs because the conventional imaging process does not make a sufficient number of signal measurements or samples. Because of this undersampling the anatomical structures outside of the FOV produce signals with the same frequency and phase characteristics of structures within the image area. This phenomenon is known as aliasing because structures outside of the FOV take on an alias in the form of the wrong spatial-encoding characteristics.

Two techniques that can be used to eliminate wraparound artifacts are illustrated in Figure 14-5. One procedure is to increase the size of the acquisition FOV and then display only the specific area of interest. The FOV is extended by increasing the number of voxels in that direction. This is described as oversampling. Under some conditions these additional samples or measurements will permit a reduction in the NSA so that acquisition time and signal-to-noise is not adversely affected by this technique.

An alternative method of eliminating wraparound or foldover artifacts is to apply presaturation pulses to the areas adjacent to the FOV. This eliminates signals and the resulting artifacts.

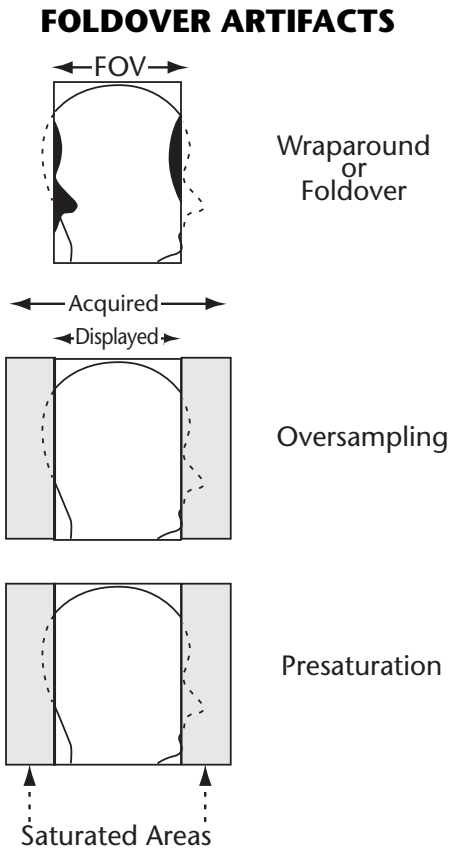


Figure 14-5. The wraparound or foldover artifact and methods for suppressing it.

Chemical-Shift Artifacts

The so-called chemical-shift artifact causes a misregistration or pixel displacement between water and fat tissue components in the frequency-encoded direction, as shown in Figure 14-6. The problem occurs because the protons in water and fat molecules do not resonate at precisely the same frequency. The shifting of the water tissue components relative to fat can produce both a void and regions of enhancement along tissue boundaries.

There are several factors that determine the number of pixels of chemical shift. Knowledge of these factors can be used to predict and control the amount of chemical shift that will appear in a clinical image.

Field Strength

We recall from Chapter 3 that the chemical shift or difference in resonant frequency between water and fat is approximately 3.3 ppm. This is the amount of chemical shift expressed as a fraction of the basic resonant frequency. The product of this and the proton resonant frequency of 64 MHz (at a field strength of 1.5 T) produces a chemical shift of 210 Hz. At a field strength of 0.5 T the chemical shift will be only 70 Hz. The practical point is that chemical shift increases with field strength and is generally more of a problem at the higher field strengths.

Bandwidth

In the frequency-encoded direction the tissue voxels emit different frequencies so that they can be separated in the reconstruction process. The RF receiver is tuned to receive this range of frequencies. This is the bandwidth of the receiver. The bandwidth is often one of the adjustable protocol factors. It can be used to control the amount of chemical shift (number of pixels), but it also has an effect on other characteristics such as signal-to-noise.

In Figure 14-6 we assume a bandwidth of 16 kHz. If the image matrix is 256 pixels in the frequency-encoded direction, this gives 15 pixels per kHz of frequency (256 pixels/16 kHz). If we now multiply this by the chemical shift of 0.210 kHz (210 Hz), we see that the chemical shift will be 3.4 pixels.

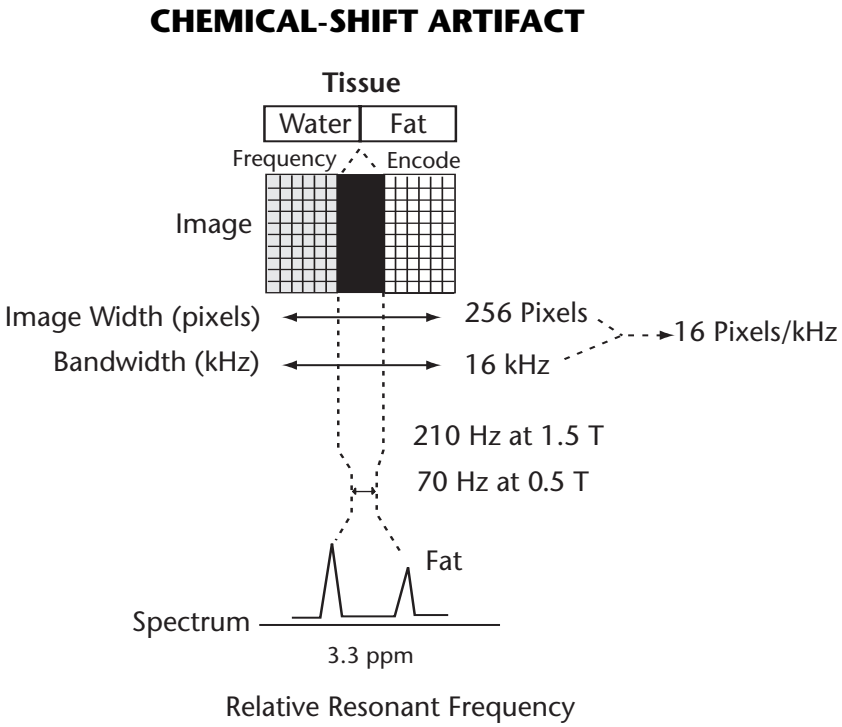


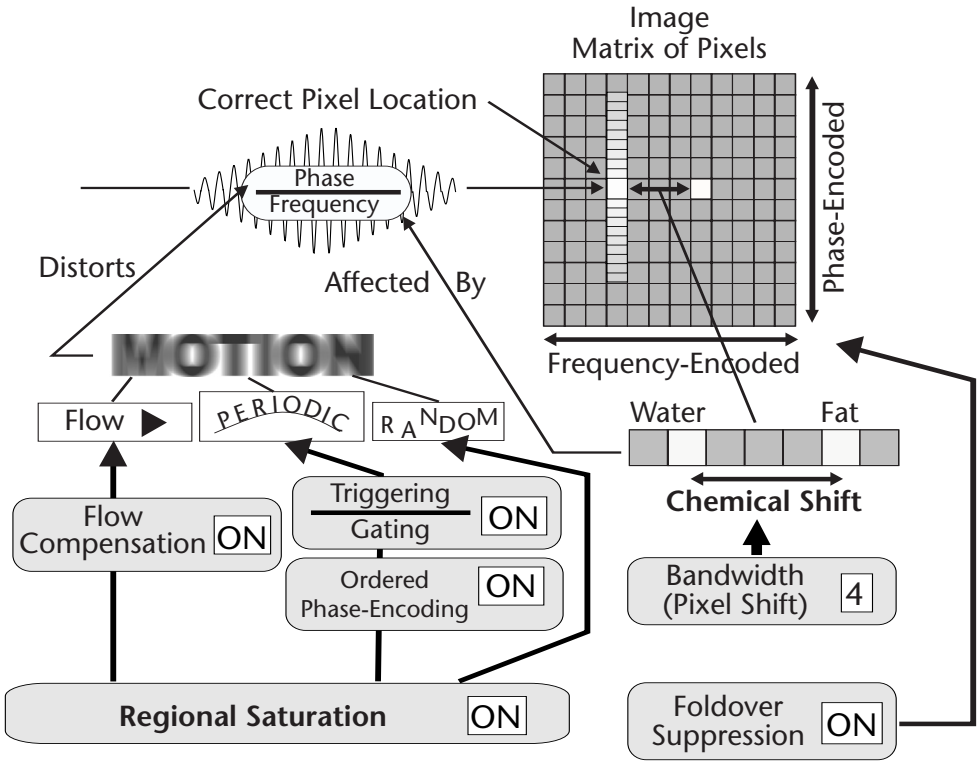
Figure 14-6. The chemical-shift artifact is related to receiver bandwidth.

The amount of chemical shift in terms of pixels can be reduced by increasing the bandwidth. This works because the chemical shift, 210 Hz, is now a smaller fraction of the image width and number of pixels.

On most MRI systems the water-fat chemical shift (number of pixels) is one of the protocol factors that can be adjusted by the operator. On some systems it is designated as bandwidth. When a different value is selected the bandwidth is automatically changed to produce the desired shift. Even though the chemical-shift artifact can be reduced by using a large bandwidth, this is not always desirable. When the bandwidth is increased, more RF noise energy will be picked up from the patient's body and the signal-to-noise relationship will be decreased. Therefore, a bandwidth or chemical-shift value should be selected that provides a proper balance between the amount of artifact and adequate signal-to-noise.

Fat suppression is useful for reducing the chemical shift artifact.

Mind Map Summary
Image Artifacts



There are a variety of artifacts that can appear in MR images. Most artifacts occur when the signal from a specific tissue voxel is not displayed in the correct image pixel location. It is possible for the signal to be displaced in both the phase-encoded and frequency-encoded directions.

Motion of the tissue or fluid produces streaks or ghost images of a pixel in the phase-encoded direction. This occurs because when motion is present, a specific tissue voxel is in different locations during the phase-encoding process from one imaging cycle to another. This produces errors in the phase encoding. We can think of signals as being given an incorrect phase address. There are several techniques that can be turned on during an image acquisition to reduce the motion artifacts.

The chemical shift artifact causes signals from fat to be displaced or shifted with respect to the tissue water signals in the frequency-encoded direction. This happens because the protons in water and fat resonate at slightly different frequencies. The amount of shift (number of pixels) in an image can be controlled by adjusting the RF receiver bandwidth. However, increasing the bandwidth to reduce the chemical shift artifact results in an increase in image noise.

Foldover is a type of artifact that can occur when the anatomical region is larger than the imaged region. The image of the tissue outside the imaged area can be folded over and appear as an artifact in the image. There are techniques that can be turned on to suppress this type of artifact.

15

MRI Safety

Introduction And Overview

MRI is not an inherently dangerous process for either the patient or the MRI staff. However, the procedure does produce a physical environment and uses forms of energy that can produce injury or discomfort if not properly controlled. The potential sources of injury are not direct biological effects but interactions of the magnetic field with other objects, which in turn, might produce injury, and the application of RF energy to the patient's body, which produces heat. During an imaging procedure a patient is subjected to:

- A strong magnetic field
- Magnetic field gradients (that are rapidly changing with time)
- RF energy
- Acoustic noise (produced by the gradients).

There are potential hazards associated with each of these if certain levels are exceeded or certain conditions exist.

The purpose of this chapter is to provide the MRI staff with an understanding of the conditions that can produce injury or discomfort and to identify the actions to take (safety procedures) to produce a safe and comfortable examination.

Magnetic Fields

As we have learned, a patient must be placed in a strong magnetic field in order to perform the imaging procedure. The magnetic field is capable of producing several effects that can

lead to unsafe conditions. Before considering the potential hazards, let us review the basic characteristics of a magnetic field.

Physical Characteristics

The general characteristics of a magnetic field were described in Chapter 2. Here we will review those characteristics that relate to safety.

The strength of a magnetic field is a characteristic that must be considered. Most clinical imaging is performed with magnetic fields in the general range of 0.15 T to 1.5 T, but there are some magnets with higher field strengths now being used for research and for some clinical imaging applications. As we will see later, certain safety-related effects are dependent on field strength.

Let us recall that there are two general areas of a magnetic field. One is the strong and relatively uniform field area within the bore of the magnet where the patient is located. The other is the external and somewhat weaker field that surrounds the magnet. This external field varies in strength with location. It is strongest where it meets the internal field and gradually becomes weaker with increased distance from the magnet. This variation in strength is a natural gradient in the field. There is no precise point at which the field ends. For all safety related purposes, the location at which the field strength drops to 5 gauss (the 5 gauss line) is considered to be the outer boundary of the field. Undesirable effects can be produced in both the internal and external field areas.

Magnetic fields can produce both mechanical and electrical effects on objects or materials located in the field. Both are potential sources of unsafe conditions.

Biological Effects

A major question in MRI today concerns the potential of undesirable effects of a magnetic field on the human body. At this time, there are no indications of any irreversible *biological effects* produced by the magnetic fields used for general clinical imaging. However, in many minds there is not a feeling of complete safety. A magnetic field, like x-radiation, is an invisible environment that carries with it a certain mystique in relation to biological effects. In the early days of x-ray imaging (over a century ago), a false sense of security led to unsuspected injury to persons receiving high levels of exposure. It has taken many years for us to develop a reasonably good understanding of the effects of x-ray and other forms of ionizing radiation. It is this experience that causes many to ask: How safe is exposure to magnetic fields? Are there unknown biological effects from a magnetic field that are going to appear sometime in the future?

First, we must recognize that there is a major physical difference between magnetic fields and ionizing radiation. A static magnetic field is not a form of energy that is directly transferred to human tissue like the various forms of radiation. We can think of it more as an environment somewhat like gravity. There is no basis for assuming that the undesirable effects produced by exposure to high levels of ionizing radiations (such as burns, mutations, and cancer-induction) will also apply to magnetic fields.

Magnetic fields, like gravity, do produce some effects, although they are distinctly different from the effects produced by radiation. The observed short-term effects of magnetic fields have occurred at relatively high field strengths beyond the general range of most clinical systems.

Biological effects arising from physical agents generally are of two forms: deterministic and stochastic. Deterministic effects are those that generally occur and are often observable at or shortly after the exposure; it is the severity of the effect that increases with the strength of the

physical agent. Burns would be an example. A stochastic effect is one that is not necessarily produced by every exposure but where there is a statistical probability of the effect occurring. Generally, it is the probability or the risk of the effect occurring that increases with the level of exposure. The induction of mutations and cancer are the two primary stochastic effects associated with exposure to ionizing radiation. These effects have not been demonstrated to occur with exposure to magnetic fields.

Internal Objects

If a patient's body contains internal ferromagnetic objects, there is a potential for injury when the patient is placed within the magnetic field. As we know, a magnetic field exerts a force or torque on ferromagnetic objects. The pulling and twisting on objects within the body can tear tissue, rupture blood vessels, and produce fatal injuries.

The most common types of objects that must be considered are the medically implanted ones such as prosthetic devices and surgical clips. In addition, there is the possibility of a person having embedded bullets, shrapnel from war wounds, swallowed objects, or pieces of metal embedded during an accident. Even very small metal fragments or particles can produce serious injury if they are located in a sensitive site such as the eye. There is a well-known case of a metal worker who lost his sight because of a small metal fragment that was under his eyelid and ruptured his eyeball when he was placed in a magnetic field for an imaging procedure.

External Objects

Ferromagnetic objects that are brought into the external magnetic field will experience a force that either attracts them to the sides of the magnet or propels them into the bore.

Safety Procedures

The principal safety procedure to safeguard against injury from internal objects is to screen patients by interview and review of medical records to determine if there are internal objects from prior surgeries or injuries.

The next step is to determine if any detected objects are potential hazards. This is sometimes difficult. The only assurance is when a specific type of device or object (such as a surgical clip) has been evaluated and found to be safe. Current information on many such devices is generally available on the World Wide Web.

Objects are pulled along the natural field gradients, from the weak to the strong areas of the field. This produces the so-called projectile effect where the objects are accelerated to high velocities and can injure anyone in their path, typically the patient.

This can occur with objects of all sizes. Accidents have occurred with large objects such as forklift tongs, metal chairs, and mop buckets. Small objects such as medical instruments carried in pockets and hairpins can easily become flying projectiles.

Magneto-electrical Effects

One of the laws of physics is that an electrical current will be induced or generated in a conducting material that is moved through a magnetic field, or is located in a changing magnetic field. In fact, this is the principle of the electric generator and the transformer. There is a possibility that motion within the body, such as cardiac activity, will generate some internal electrical currents. However,

Safety Procedures

The principal safety procedure is proper training of all persons who enter the room containing the magnet.

Signs should be posted informing patients and staff about the presence of a strong magnetic field and the associated hazards.

The imaging staff should monitor all other persons who enter the room and who might bring hazardous objects into the room.

The magnet room should be properly secured when not under the direct observation and supervision of the imaging staff.

Special attention should be given to ensure the safety of cleaning and maintenance staff who might not have adequate safety training and who often work during non-operational hours with minimum supervision.

these currents do not appear to produce any undesirable effects, such as fibrillation, at the field strengths used for clinical imaging.

The most common effect of this type is an elevated T wave in the EKG signal from a patient being monitored in the magnetic field. This is the time within the cardiac cycle when the blood flow velocity is the highest. A voltage is generated by this flow through the magnetic field that adds to the normal T wave, making it appear to be enhanced. It does not represent a change in cardiac activity.

Activation of or Damage to Implanted Devices

Some implanted electronic devices, such as cardiac pacemakers, have magnetically activated switches. The reason for this is so physicians can temporarily turn off the pacemaker for testing by placing a small magnet on the surface of the patient near the pacemaker site. The concern is that if a patient with this type of pacemaker should enter a magnetic field, the pacemaker would be turned off, which could be potentially fatal. This type of activation could also occur in the external field surrounding a magnet. This is why it is a common practice to exclude patients with pacemakers from the field areas within the 5 gauss line unless it has been determined that the field will not have an adverse effect on their device.

There is a possibility that some implanted electronic devices could be damaged by the high field strength within the magnet.

Safety Procedures

Exclude persons with implanted electronic devices from the magnetic field unless it has been determined that there is no adverse effect on a particular device.

Gradients

We recall that a gradient is a non-uniformity in the magnetic field produced by turning on the various gradient coils. A gradient is a variation in field strength over space, which in itself is not a safety problem. However, potential problems are produced at the times when the gradients are being turned on and off. During these times the magnetic field is changing rapidly with respect to time. A time-varying magnetic field induces or generates electrical

Safety Procedures

It is assumed that gradient design is within appropriate safety limits.

currents through conductive materials located within the changing field, as illustrated in Figure 15-1. This includes human tissue as well as any implanted materials or devices.

For a specific material or device the amount of current produced depends on the rate at which the magnetic field is changed. This is expressed as the factor, dB/dt, where B is the symbol for field strength. This is in the units of tesla per second. The U.S. Food and Drug Administration (FDA) provides guidelines on dB/dt values that can be considered to be safe operating levels. A maximum value of 6 T/sec applies to an imaging system, but other, higher values apply to specific imaging conditions.

All MR systems do not have gradients that change at the same rate (dB/dt). This is actually the gradient slew rate described in Chapter 2. The trend in equipment development is to go to higher values to perform certain types of acquisitions, such as echo planar imaging.

RF Energy

During an imaging procedure pulses of RF energy are transmitted to the patient's body. Most of the energy is absorbed by the tissue and is converted to heat that has the potential of increasing the temperature within the body. This is the physical principle of the microwave oven. In the microwave oven high-frequency radio energy (microwaves) are transmitted at a high power level to a relatively small mass of food. This high concentration of energy can raise the temperature by hundreds of degrees.

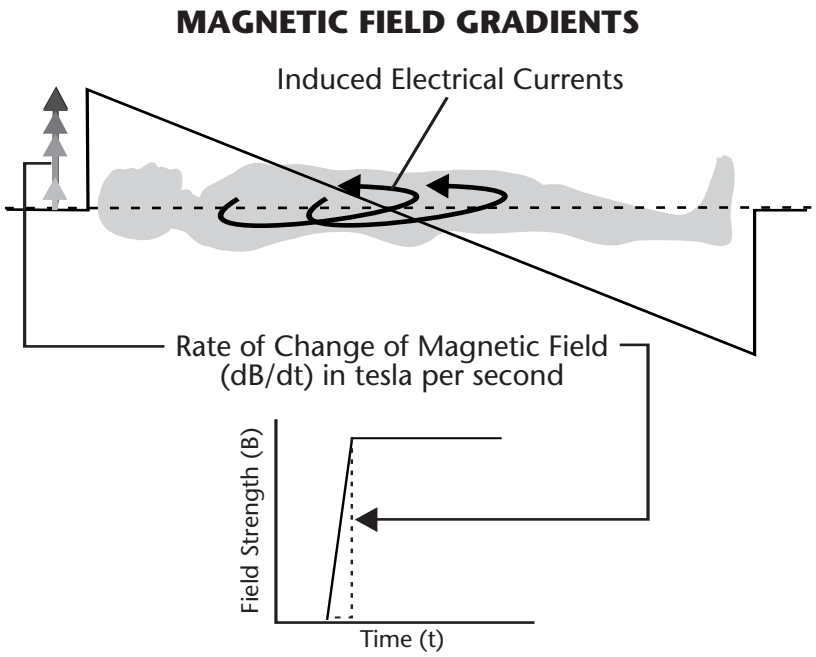


Figure 15-1. Electrical currents are generated by rapidly changing magnetic fields (the gradients).

In MRI the RF energy applied to a patient's body is at a much lower power and is distributed over a larger mass of tissue than in the microwave oven example. The elevation in temperature produced during MRI depends on many factors. Variations among patients that will affect this include body size and blood circulation, which distributes heat and promotes cooling.

The principal issue associated with an imaging procedure is the rate at which the energy is transferred to the body. The factors that determine this are illustrated in Figure 15-2. Recall that the rate of energy transfer is the physical quantity *power* and is expressed in the units of *watts*. The rate at which temperature is increased is related to the concentration of power in tissue.

Specific Absorption Rate (SAR)

In MRI, the rate at which energy (heat) is deposited in a unit mass of tissue is expressed in terms of the *Specific Absorption Rate* (SAR) in units of watts per kilogram of tissue. For specific tissue conditions, the rate of temperature rise will be proportional to the SAR. The total increase in temperature is determined by the SAR, the duration of the exposure to the RF pulses, and various tissue cooling and heat distribution factors.

Limits have been established that are thought to represent insignificant risk to patients if the examination times are within established ranges. These are shown in Figure 15-2.

The SAR is determined by a combination of factors associated with the MR system and

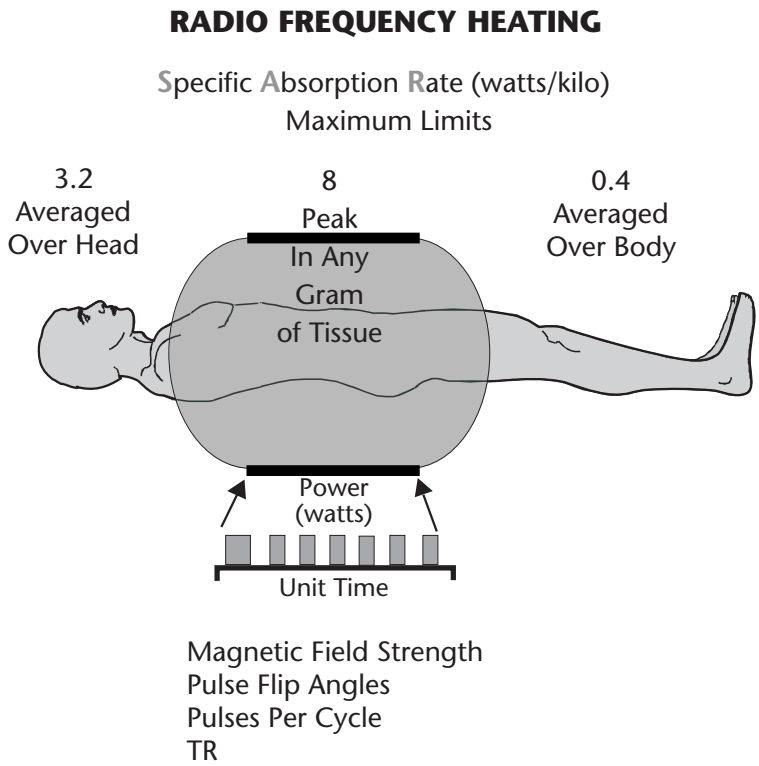


Figure 15-2. The factors that determine the SAR, or rate of heat production during an image acquisition.

the acquisition protocol as listed in Figure 15-2. Each RF pulse delivers a quantity of energy to the body. Therefore, the SAR is determined by the energy in each pulse and the rate at which the pulses are applied.

Magnetic Field Strength

The field strength is a factor because the amount of RF energy required to produce a pulse with a specific flip angle increases with field strength. Therefore, a 90° pulse delivers more energy in a 1.5 T field than in a 0.5 T field.

Pulse Flip Angles

The energy of a pulse increases with flip angle. A 180° pulse delivers more energy than a 90° pulse.

Pulses per Cycle

We have observed that there are many combinations of RF pulses used by the different imaging methods, or pulse sequences. A simple spin echo acquisition might use just two pulses (90° and 180°), while other methods, such as fast or turbo spin echo acquisitions use many 180° pulses. This produces more energy delivered during each imaging cycle.

TR

Since TR is the duration of the imaging cycle, it affects the rate at which energy is delivered. Reducing TR concentrates the pulses into a shorter time and results in an increase in the SAR.

Determining the SAR for a Patient

We have just seen that there are many variable factors that determine the SAR for a specific patient undergoing an imaging procedure. These include both the size of the patient and the specific imaging protocol that is being used. It would be difficult to take all of the factors and manually calculate the SAR value. Fortunately, most MR systems perform this calculation during the preparation of the patient and of the protocol. The SAR value might be displayed for the operator to visually check and to automatically prevent the system from proceeding with an imaging protocol that will exceed established SAR limits.

Surface Burns

It is possible for metal objects, such as monitoring leads and electrodes that are in contact with the patient's body to act as an antenna and to pick up some of the RF energy. There have been cases where the concentration of energy at the point of contact has produced burns.

Safety Procedures

Be aware of the potential problem and use only devices and lead configurations that have been demonstrated to be safe. Investigate any reports from the patient of discomfort.

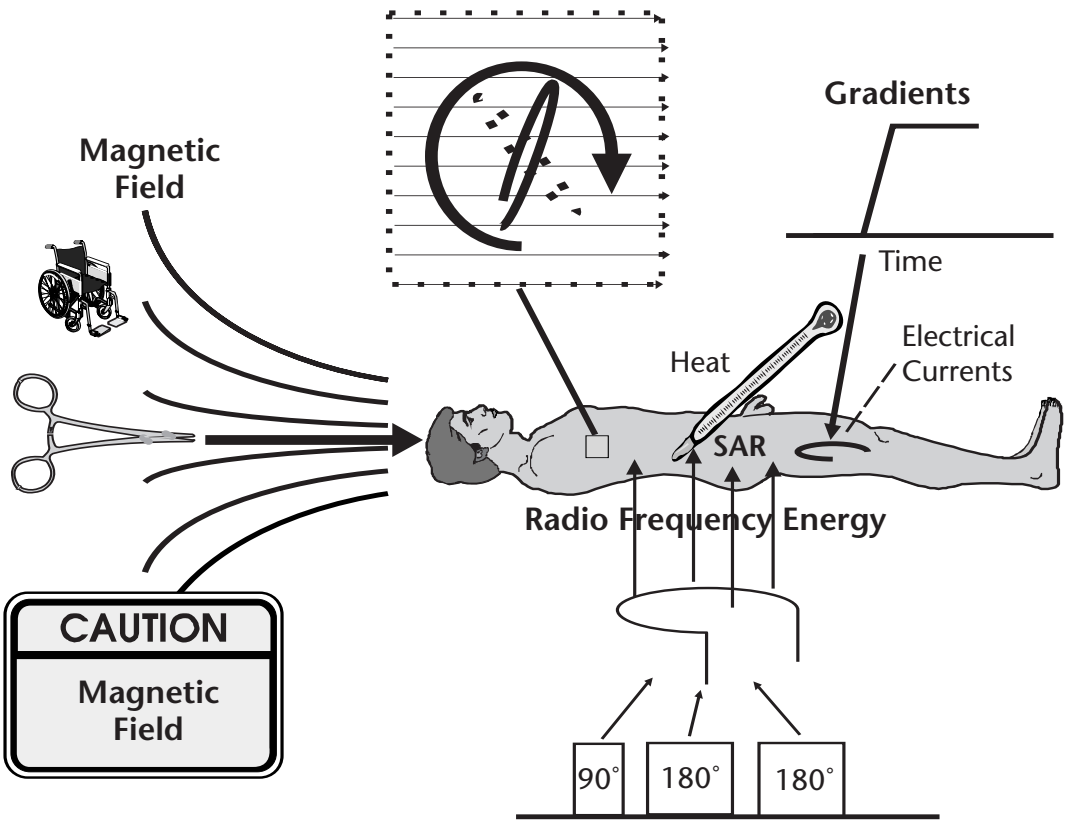
Acoustic Noise

The noise produced by the gradients is loud, especially for high performance gradients performing certain types of acquisitions. This is often uncomfortable for patients, and can be a potential source of longer term effects to hearing.

Safety Procedures

Provide hearing protection and audio diversions as appropriate.

Mind Map Summary
MRI Safety



The MR imaging procedure is generally safe for both patients and staff if certain precautions are observed. However, there are three, specific physical environments that, if not properly controlled, could produce injury. They are: a strong magnetic field, rapid-changing magnetic field gradients, and RF energy.

The strong static (non-changing) magnetic field does not appear to produce any undesirable direct biological effects. However, injury can result from the effect of the magnetic field on metal objects, both external to and within the patient's body. Objects, such as surgical clips and other implanted devices, can be pulled or rotated by the magnetic field. Magnetic-susceptible objects brought into the external field can be pulled into the magnetic field and become projectiles that can injure patients.

A gradient, that is a rapidly changing magnetic field when it is being turned on and off, can induce electrical currents within a patient's body or equipment attached to the patient. This is why limits have been established on the rate of change for the gradient fields.

The RF energy applied to a patient's body is absorbed and converted to heat. This can increase tissue temperatures. The SAR is the quantity that expresses the rate at which energy is being deposited. SAR is determined by several factors: the number of pulses in a time interval, the flip angles of the pulses, and the strength of the magnetic field. Limits have been established to minimize adverse effects.

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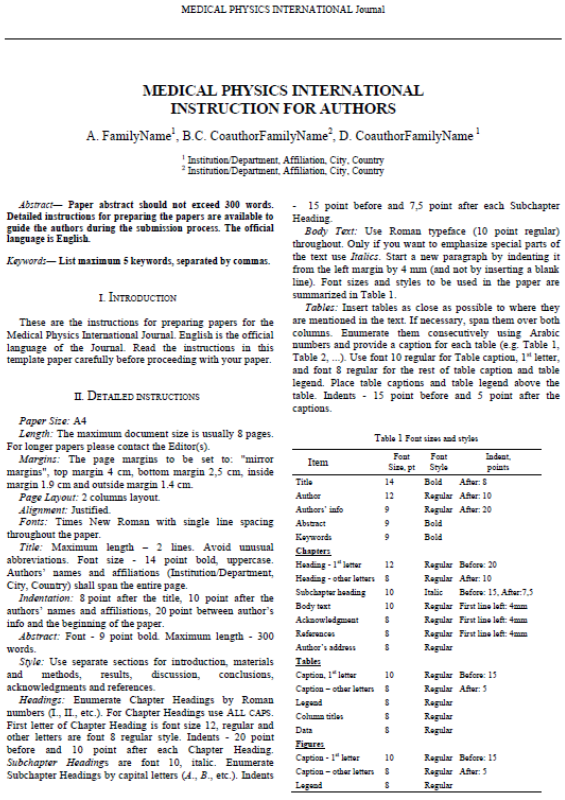

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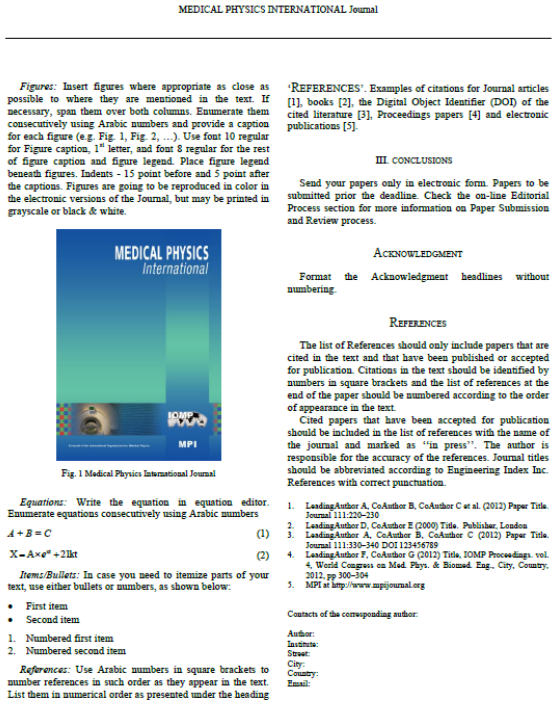

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2. LeadingAuthor D, CoAuthor E (2000) Title. Publisher, London
3. LeadingAuthor A, CoAuthor B, CoAuthor C (2012) Paper Title. Journal 111:330–340 DOI 123456789
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