Provides a comprehensive overview of the basic concepts behind the application and designs of medical instrumentation

This premiere reference on medical instrumentation describes the principles, applications, and design of the medical instrumentation most commonly used in hospitals. It places great emphasis on design principles so that scientists with limited background in electronics can gain enough information to design instruments that may not be commercially available. The revised edition includes new material on microcontroller-based medical instrumentation with an example of ECG-based circuit schematics and relevant code, device design with circuit simulations and implementations, dry electrodes for electrocardiography and medical imaging techniques. Chapters include new problems and updated reference material that covers the latest medical technologies.

Medical Instrumentation: Application and Design, Fifth Edition covers general concepts that are applicable to all instrumentation systems, including the static and dynamic characteristics of a system, the engineering design process, the commercial development and regulatory classifications, and the electrical safety, protection, codes and standards for medical devices. The readers learn about the principles behind various sensor mechanisms, the necessary amplifier and filter designs for analog signal processing, and the digital data acquisition, processing, storage and display using microcontrollers. The measurements of both cardiovascular dynamics and respiratory dynamics are discussed, as is the developing field of biosensors. The book also covers general concepts of clinical laboratory instrumentation, medical imaging, various therapeutic and prosthetic devices, and more.

- Emphasizes design throughout so scientists and engineers can create medical instruments
- Updates the coverage of modern sensor signal processing
- New material added to the chapter on modern microcontroller use
- Features revised chapters, descriptions, and references throughout
- Includes many new worked out examples and supports student problem-solving
- Offers updated, new, and expanded materials on a companion webpage
- Supplemented with a solutions manual containing complete solutions to all problems

Medical Instrumentation: Application and Design, Fifth Edition is an excellent book for a senior to graduate-level course in biomedical engineering and will benefit other health professionals involved with the topic.

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APPLICATION AND DESIGN

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MEDICAL INSTRUMENTATION

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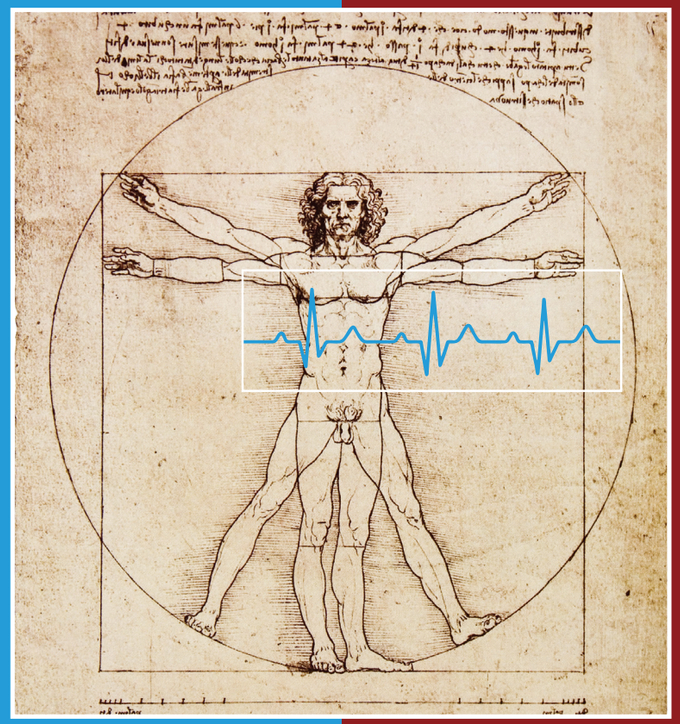

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EDITED BY
JOHN G. WEBSTER
AMIT J. NIMUNKAR

FIFTH EDITION

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MEDICAL INSTRUMENTATION

APPLICATION AND DESIGN

MEDICAL INSTRUMENTATION

Application and Design

FIFTH EDITION

Edited by

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PREFACE

Medical Instrumentation: Application and Design, Fifth Edition, is written for a senior to graduate-level course in biomedical engineering. It describes the principles, applications, and design of the medical instruments most commonly used in hospitals. Because equipment changes with time, we have stressed fundamental principles of operation and general types of equipment, avoiding detailed descriptions and photographs of specific models. Furthermore, because biomedical engineering is an interdisciplinary field, requiring good communication with health-care personnel, we have provided some applications for each type of instrument. However, to keep the book to a reasonable length, we have omitted much of the physiology.

Most of those who use this text have had an introductory course in chemistry, are familiar with mathematics through differential equations, have a strong background in physics, and have taken courses in electric circuits and electronics. However, readers without this background will gain much from the descriptive material and should find this text a valuable reference. In addition, we recommend reading background material from an inexpensive physiology text, such as W. F. Ganong's *Review of Medical Physiology*, 26th edition (New York: McGraw-Hill Education, 2019).

EMPHASIS ON DESIGN

Throughout the book, we emphasize design. A scientist or engineer who has some background in electronics and instrumentation will glean enough information, in many of the areas we address, to be able to design medical instruments. This ability should be especially valuable in those situations—so frequently encountered—where special instruments that are not commercially available are required.

PEDAGOGY

The book provides 282 problems, located at the end of each chapter, plus 127 in-text worked examples. Problems are designed to cover a wide variety of applications ranging from analysis of the waves of the electrocardiogram to circuit design of biopotential amplifiers with microcontroller implementation and identification of electric safety hazards.

REFERENCES

Rather than giving an exhaustive list of references, we have provided a list of review articles and books that can serve as a point of departure for further study on any given topic.

ORGANIZATION

Each chapter has been carefully reviewed and updated for the fifth edition, and many new problems and references are included.

Chapter 1 covers general concepts that are applicable to all instrumentation systems, including the commercial development of medical instruments, on biostatistics, and on the regulation of medical devices, and the design of amplifiers. Chapter 2 describes basic sensors, and Chapter 3 presents microcontroller implementation in medical devices. Chapters 4 through 6 deal with biopotentials, tracing the topic from the origin of biopotentials, through electrodes, to the special amplifier design required.

Chapters 7 and 8 cover the measurement of cardiovascular dynamics–pressure, sound, flow, and volume of blood. Chapter 9 presents the measurement of respiratory dynamics–pressure, flow, and concentration of gases.

Chapter 10 describes the developing field of biosensors: sensors that measure chemical concentrations within the body via catheters or implants. Chapter 11 describes that area in the hospital where the greatest number of measurements are made, the clinical laboratory. Chapter 12 starts with general concepts of medical imaging and shows their applications to x-ray techniques, magnetic resonance imaging, positron emission tomography, and Doppler ultrasonic imaging.

Chapter 13 deals with devices used in therapy, such as the pacemaker, defibrillator, cochlear prosthesis, transcutaneous electrical nerve stimulation, implantable automatic defibrillators, the total artificial heart, lithotripsy, high-frequency ventilators, infant incubators, drug infusion pumps, and anesthesia machines. Chapter 14 presents a guide both to electric safety in the hospital and to minimization of hazards.

We have used the internationally recommended SI units throughout this book. In the case of units of pressure, we have presented both the commonly used millimeters of mercury and the SI unit, the pascal. To help the reader follow the trend toward employing SI units, the Appendix provides the most common conversion factors. The Appendix also provides a number of physical constants used in the book and a list of abbreviations.

A Solutions Manual containing complete solutions to all problems is available for the instructors at www.wiley.com/go/Webster/Medical instrumentation5e

LIST OF SYMBOLS

This list gives single-letter symbols for quantities, without subscripts or modifiers. Symbols for physical constants are given in Appendix A.1, multi-letter symbols in Appendix A.4, and chemical symbols in Appendix A.5.

Symbol	Quantity	Introduced in Section
a	Absorptivity	10.3
a	Activity	5.2
a	Coefficient	1.10
a	Lead vector	6.2
A	Absorbance	10.3
A	Area	2.2
A	Coefficient	1.10
A	Gain	1.11
A	Magnetic vector potential	4.9
A	Percent	1.9
b	Coefficient	1.10
b	Intercept	1.9
B	Coefficient	1.10
B	Percent	1.9
B	Viscous friction	1.10
B	Magnetic flux density	8.3
c	Coefficient	7.11
c	Specific heat	8.2
c	Velocity of sound	8.4
C	Capacitance	1.10
C	Compliance	1.10
C	Concentration	8.1
C	Contrast	12.1
d	Diameter	5.9
d	Distance	4.1
d	Duration	14.2
D	*d/dt*	1.10

Symbol	Quantity	Introduced in Section
D	Detector responsivity	2.19
D	Diameter	5.9
D	Diffusing capacity	9.8
D	Digital signal	3.4
D	Distance	4.4
E	emf	2.10
E	Energy	2.15
E	Irradiance	2.19
E	Modulus of elasticity	1.10
f	Force	2.7
f	Frequency	1.18
F	Faraday constant	4.1
F	Filter transmission	2.19
F	Flow rate	7.3
F	Force	7.14
F	Molar fraction	9.3
g	Conductance/area	4.1
g	Gravity acceleration	7.11
G	Form factor	2.4
G	Gage factor	2.2
G	Gain	1.14
G	Transfer function	1.7
h	Height	7.11
H	Feedback gain	1.7
i	Current	1.12
I	Current	1.9
I	Intensity	12.5
j	$+\sqrt{-1}$	1.10
J	Current density	4.9
J	Number of standard deviations	12.1
k	Constant	6.8
k	Piezoelectric constant	2.7
K	Constant	1.10
K	Number	12.1
K	Sensitivity	1.10
K	Solubility product	5.3
K	Spring constant	1.10
L	Inductance	2.4
L	Inertance	7.3
L	Length	2.2
m	Average number	12.1
m	Mass	7.3
m	Slope	1.9
M	Mass	1.10

(*Continued*)

Symbol	Quantity	Introduced in Section
M	Measured values	12.2
M	Modulation	12.1
$\mathbf{M}$	Cardiac vector	6.2
n	Number	1.8
n	Refractive index	2.16
n	Valence	4.1
N	Noise equivalent bandwidth	12.3
N	Number	5.3
N	Numerical aperture	7.1
N	Turns ratio	1.23
p	Change in pressure	9.1
p	Probability	12.1
P	Permeability	4.1
P	Power	1.9
P	Pressure	7.3
P	Projection	12.7
q	Change in volume flow	9.1
q	Charge	2.7
Q	Heat content	8.2
Q	Volume flow	9.1
r	Correlation coefficient	1.8
r	Radius	7.3
r	Resistance per length	4.1
R	Range	8.4
R	Impedance	1.11
R	Ratio	10.3
R	Resistance	1.10
R	Universal gas constant	4.1
s	d/dt	1.10
s	Standard deviation	1.8
S	Modulation transfer function	12.2
S	Saturation	10.1
S	Slew rate	1.21
S	Source output	2.19
t	Thickness	5.9
t	Time	1.10
T	Temperature	2.10
T	Tensile force	7.14
T	Time	3.4
T	Transmittance	11.1
u	Velocity	4.4
u	Work function	12.5
U	Molar uptake	9.1
v	Voltage	1.11
v	Change in volume	9.1

Symbol	Quantity	Introduced in Section
V	Voltage	1.10
V	Volume	7.3
W	Radiant Power	1.12
W	Power	8.5
W	Weight	10.3
W	Weighting factor	12.7
x	Constant	10.3
x	Distance	2.6
x	Input	1.7
X	Chemical species	9.1
X	Effort variable	1.9
X	Flow variable	1.9
X	Value	1.8
y	Constant	10.3
y	Output	1.7
Y	Admittance	1.9
Y	Flow variable	1.9
Y	Value	1.8
z	Distance	4.1
Z	Atomic number	12.5
Z	Impedance	1.6

Greek Letters

Symbol	Quantity	Introduced in Section
α	Polytropic constant	9.5
α	Temperature coefficient	2.11
α	Thermoelectric sensitivity	2.10
β	Material constant for thermistor	2.11
γ	Gyromagnetic ratio	12.8
Δ	Deviation	10.3
ε	Dielectric constant	2.6
ε	Emissivity	2.12
ξ	Damping ratio	1.10
η	Viscosity	1.10
θ	Angle	2.16
Λ	Logarithmic decrement	1.10
λ	Wavelength	2.12
μ	Absorption coefficient	12.7
μ	Mobility	5.2
μ	Permeability	2.4
μ	Poisson's ratio	2.2

(*Continued*)

Symbol	Quantity	Introduced in Section
υ	Frequency	2.15
ρ	Density	1.10
ρ	Mole density	9.1
ρ	Resistivity	2.2
σ	Electrical conductivity	13.4
σ	Stefan–Boltzmann constant	12.5
τ	Time constant	1.10
ϕ	Divergence	8.4
ϕ	Phase shift	1.10
Φ	Potential	4.6
ω	Frequency	1.10

CONTENTS

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The authors welcome your suggestions for improvement of subsequent printings and editions.

John G. Webster
Amit J. Nimunkar

1

BASIC CONCEPTS OF MEDICAL INSTRUMENTATION

Walter H. Olson and John G. Webster

The invention, prototype design, product development, clinical testing, regulatory approval, manufacturing, marketing, and sale of a new medical instrument add up to a complex, expensive, and lengthy process. Very few new ideas survive the practical requirements, human barriers, and inevitable setbacks of this arduous process. Usually there is one person who is the "champion" of a truly new medical instrument or device. This person—who is not necessarily the inventor—must have a clear vision of the final new product and exactly how it will be used. And most important, this person must have the commitment and persistence to overcome unexpected technical problems, convince the naysayers, and cope with the bureaucratic apparatus that is genuinely needed to protect patients.

Important new ideas rarely flow smoothly to widespread clinical use. There are probably one hundred untold failure stories for each success story! New inventions usually are made by the wrong person with the wrong contacts and experience, in the wrong place at the wrong time. It is important to understand the difference between a crude feasibility prototype and a well-developed, reliable, manufacturable product. Patents are important to protect ideas during the development process and to provide incentives for making the financial investments needed. Many devices have failed because they were too hard to use, reliability and ruggedness were inadequate, marketing was misdirected, user education was lacking, or service was poor and/or slow.

An evolutionary product is a new model of an existing product that adds new features, improves the technology, and reduces the cost of production. A revolutionary new product either solves a totally new problem or uses

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a new principle or concept to solve an old problem in a better way that displaces old methods. A medical instrument that improves screening, diagnosis, or monitoring may not add value by improving patient outcome unless improvements in the application of therapy occur as a result of using the medical instrument.

1.1 TERMINOLOGY OF MEDICINE AND MEDICAL DEVICES

Most biomedical engineers learn the physical sciences first in the context of traditional engineering, physics, or chemistry. When they become interested in medicine, they usually take at least a basic course in physiology, which does not describe disease or pathologic terminology. The book *Medical Terminology: An Illustrated Guide* (Cohen and DePetris, 2013) is recommended. It emphasizes the Latin and Greek roots in a workbook format (with answers) that includes clinical case studies, flash cards, and simple, clear illustrations. An unabridged medical dictionary such as *Dorland's Illustrated Medical Dictionary*, 30th ed. (Dorland, 2003), is often useful. Physicians frequently use abbreviations and acronyms that are difficult to look up, and ambiguity or errors result. Six references on medical abbreviations are given (Cohen and DePetris, 2013; Davis, 2011; Firkin and Whitworth, 2001; Haber, 1988; Hamilton and Guides, 1988; Heister, 1989). Medical eponyms are widely used to describe diseases and syndromes by the name of the person who first identified them. Refer to *Dictionary of Medical Eponyms* (Firkin and Whitworth, 2001).

The name used to describe a medical instrument or device should be informative, consistent, and brief. The annual *Health Devices Sourcebook* (Anonymous, 2013a) is a directory of U.S. and Canadian medical device products, trade names, manufacturers, and related services. This book uses internationally accepted nomenclature and a numerical coding system for over 5000 product categories. The *Product Development Directory* (Anonymous, 1996) lists all specific medical products by the FDA standard product category name since enactment of the Medical Devices Amendments in April 1976. The *Encyclopedia of Medical Devices and Instrumentation*, 2nd edition (Webster, 2006), vols. 1–6 has many detailed descriptions. But beware of borrowing medical terminology to describe technical aspects of devices or instruments. Confounding ambiguities can result.

Recent information on medical instrumentation can be found by searching World Wide Web servers such as www.fda.gov, www.uspto.gov, Library Online Catalogs, and journal electronic databases such as Engineering Village, Science Citation Index, and PubMed.

1.2 GENERALIZED MEDICAL INSTRUMENTATION SYSTEM

Every instrumentation system has at least some of the functional components shown in Figure 1.1. The primary flow of information is from left to right. Elements and relationships depicted by dashed lines are not essential. The major difference between this system of medical instrumentation and conventional instrumentation systems is that the source of the signals is living tissue or energy applied to living tissue.

MEASURAND

The physical quantity, property, or condition that the system measures is called the measurand. The accessibility of the measurand is important because it may be internal (blood pressure), it may be on the body surface (electrocardiogram (ECG) potential), it may emanate from the body (infrared radiation), or it may be derived from a tissue sample (such as blood or a biopsy) that is removed from the body. Most medically important measurands can be grouped in the following categories: biopotential, pressure, flow, dimensions (imaging), displacement (velocity, acceleration, and force),

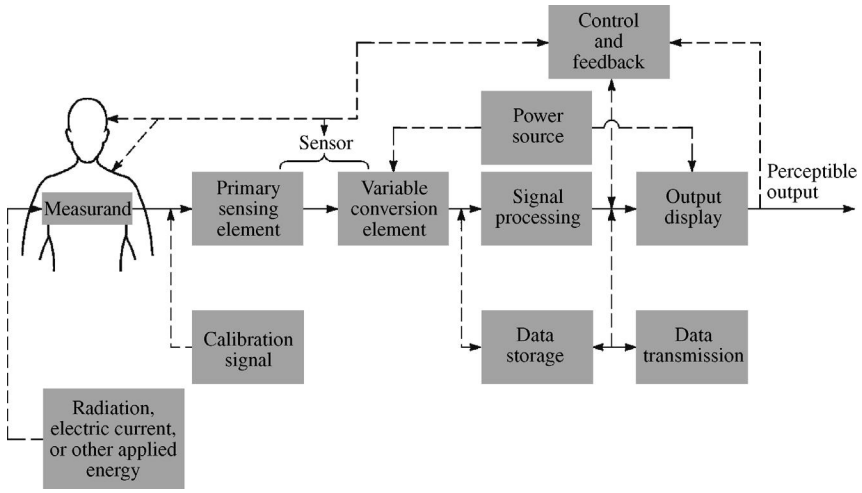


Figure 1.1 Generalized instrumentation system The sensor converts energy or information from the measurand to another form (usually electric). This signal is then processed and displayed so that humans can perceive the information. Elements and connections shown by dashed lines are optional for some applications.

impedance, temperature, and chemical concentrations. The measurand may be localized to a specific organ or anatomical structure.

SENSOR

Generally, the term *transducer* is defined as a device that converts one form of energy to another. A sensor converts a physical measurand to an electric output. The sensor should respond only to the form of energy present in the measurand, to the exclusion of all others. The sensor should interface with the living system in a way that minimizes the energy extracted, while being minimally invasive. Many sensors have a primary sensing element such as a diaphragm, which converts pressure to displacement. A variable-conversion element, such as a strain gage, then converts displacement to an electric voltage. Sometimes the sensitivity of the sensor can be adjusted over a wide range by altering the primary sensing element. Many variable-conversion elements need external electric power to obtain a sensor output.

SIGNAL CONDITIONING

Usually the sensor output cannot be directly coupled to the display device. Simple signal conditioners may only amplify and filter the signal or merely match the impedance of the sensor to the display. Often sensor outputs are converted to digital form and then processed by a microcontroller (Tompkins and Webster, 1981). For example, signal filtering may reduce undesirable sensor signals. It may also average repetitive signals to reduce noise, or it may convert information from the time domain to the frequency domain.

OUTPUT DISPLAY

The results of the measurement process must be displayed in a form that the human operator can perceive. The best form for the display may be numerical or graphical, discrete or continuous, permanent or temporary—depending on the particular measurand and how the operator will use the information. Although most displays rely on our visual sense, some information (Doppler ultrasonic signals, for example) is best perceived by other senses (here, the auditory sense). User controls and the output display should conform to the *Human Factors Engineering Guidelines and Preferred Practices for the Design of Medical Devices* [ANSI/AAMI, 2009/R(2013)] (Anonymous, 2013b).

AUXILIARY ELEMENTS

A calibration signal with the properties of the measurand should be applied to the sensor input or as early in the signal-processing chain as possible. Many forms of control and feedback may be required to elicit the measurand, to adjust the sensor and signal conditioner, and to direct the flow of output for display, storage, or transmission. Control and feedback may be automatic or manual. Data may be stored briefly to meet the requirements of signal conditioning or to enable the operator to examine data that precede alarm conditions. Or data may be stored before signal conditioning, so that different processing schemes can be utilized. The data can be transmitted wirelessly to remote displays at nurses' stations, medical centers, or medical data-processing facilities.

1.3 ALTERNATIVE OPERATIONAL MODES

DIRECT–INDIRECT MODES

Often the desired measurand can be interfaced directly to a sensor because the measurand is readily accessible or because acceptable invasive procedures are available. When the desired measurand is not accessible, we can use either another measurand that bears a known relation to the desired one or some form of energy or material that interacts with the desired measurand to generate a new measurand that is accessible. Examples include cardiac output (volume of blood pumped per minute by the heart), determined from measurements of respiration and blood gas concentration or from dye dilution; morphology of internal organs, determined from x-ray shadows; and pulmonary volumes, determined from variations in thoracic impedance plethysmography.

SAMPLING AND CONTINUOUS MODES

Some measurands such as body temperature and ion concentrations change so slowly that they may be sampled infrequently. Other quantities such as the electrocardiogram and respiratory gas flow may require continuous monitoring. The frequency content of the measurand, the objective of the measurement, the condition of the patient, and the potential liability of the physician all influence how often medical data are acquired. Many data that are collected may go unused.

GENERATING AND MODULATING SENSORS

Generating sensors produce their signal output from energy taken directly from the measurand, whereas modulating sensors use the measurand to alter the flow of energy from an external source in a way that affects the output of the sensor. For example, a photovoltaic cell is a generating sensor because it provides an output voltage related to its irradiation, without any additional external energy source. However, a photoconductive cell is a modulating sensor; to measure its change in resistance with irradiation, we must apply external energy to the sensor.

ANALOG AND DIGITAL MODES

Signals that carry measurement information are either *analog*, meaning continuous and able to take on any value within the dynamic range, or *digital*, meaning discrete and able to take on only a finite number of different values. Most currently available sensors operate in the analog mode, although some inherently digital measuring devices have been developed. Increased use of digital signal processing has required concurrent use of analog-to-digital and digital-to-analog converters to interface computers with analog sensors and analog display devices. Researchers have developed indirect digital sensors that use analog primary sensing elements and digital variable-conversion elements (optical shaft encoders). Also quasi-digital sensors, such as quartz-crystal thermometers, give outputs with variable frequency, pulse rate, or pulse duration that are easily converted to digital signals.

The advantages of the digital mode of operation include greater accuracy, repeatability, reliability, and immunity to noise. Furthermore, periodic calibration is usually not required. Digital numerical displays are replacing most analog meter movements because of their greater accuracy and readability. Most clinicians prefer digital displays when they are determining whether a physiological variable is within certain limits and when they are looking at a parameter that can change quickly, such as beat-to-beat heart rate.

REAL-TIME AND DELAYED-TIME MODES

Of course sensors must acquire signals in real time as the signals actually occur. The output of the measurement system may not display the result immediately, however, because some types of signal processing, such as averaging and transformations, need considerable input before any results can be produced. Often such short delays are acceptable unless urgent

feedback and control tasks depend on the output. In the case of some measurements, such as cell cultures, several days may be required before an output is obtained.

1.4 MEDICAL MEASUREMENT CONSTRAINTS

The medical instrumentation described throughout this book is designed to measure various medical and physiological parameters. The principal measurement and frequency ranges for each parameter are major factors that affect the design of all the instrument components shown in Figure 1.1. To get a brief overview of typical medical parameter magnitude and frequency ranges, refer to Table 1.1. Shown here are approximate ranges that are intended to include normal and abnormal values. Most of the parameter measurement ranges are quite low compared with nonmedical parameters. Note, for example, that most voltages are in the microvolt range and that pressures are low (about 100 mm Hg = 1.93 psi = 13.3 kPa). Also note that all the signals listed are in the audio frequency range or below and that many signals contain dc and very low frequencies. These general properties of medical parameters limit the practical choices available to designers for all aspects of instrument design.

Many crucial variables in living systems are inaccessible because the proper measurand–sensor interface cannot be obtained without damaging the system. Unlike many complex physical systems, a biological system is of such a nature that it is not possible to turn it off and remove parts of it during the measurement procedure. Even if interference from other physiological systems can be avoided, the physical size of many sensors prohibits the formation of a proper interface. Either such inaccessible variables must be measured indirectly, or corrections must be applied to data that are affected by the measurement process. The cardiac output is an important measurement that is obviously quite inaccessible.

Variables measured from the human body or from animals are seldom deterministic. Most measured quantities vary with time, even when all controllable factors are fixed. Many medical measurements vary widely among normal patients, even when conditions are similar. This inherent *variability* has been documented at the molecular and organ levels, and even for the whole body. Many internal anatomical variations accompany the obvious external differences among patients. Large tolerances on physiological measurements are partly the result of interactions among many physiological systems. Many feedback loops exist among physiological systems, and many of the interrelationships are poorly understood. It is seldom feasible to control or neutralize the effects of these other systems on the measured variable. The most common method of coping with this variability is to assume

Table 1.1 Medical and Physiological Parameters

Parameter or Measuring Technique	Principal Measurement Range of Parameter	Signal Frequency Range, Hz	Standard Sensor or Method
Ballistocardiography	0–7 mg	dc–40	Accelerometer, strain gage
	0–100 μm	dc–40	Displacement linear variable differential transformer
Bladder pressure	1–100 cm H_2O	dc–10	Strain-gage manometer
Blood flow	1–300 ml/s	dc–20	Flowmeter (electromagnetic or ultrasonic)
Blood pressure, arterial			
Direct	10–400 mm Hg	dc–50	Strain-gage manometer
Indirect	25–400 mm Hg	dc–60	Cuff, auscultation
Blood pressure, venous	0–50 mm Hg	dc–50	Strain gage
Blood gases			
P_{O_2}	30–100 mm Hg	dc–2	Specific electrode, volumetric or manometric
P_{CO_2}	40–100 mm Hg	dc–2	Specific electrode, volumetric or manometric
P_{N_2}	1–3 mm Hg	dc–2	Specific electrode, volumetric or manometric
P_{CO}	0.1–0.4 mm Hg	dc–2	Specific electrode, volumetric or manometric
Blood pH	6.8–7.8 pH units	dc–2	Specific electrode
Cardiac output	4–25 liter/min	dc–20	Dye dilution, Fick
Electrocardiography	0.5–4 mV	0.01–250	Skin electrodes
Electroencephalography	5–300 μV	dc–150	Scalp electrodes
Electrocorticography and brain depth	10–5000 μV	dc–150	Brain-surface or depth electrodes
Electrogastrography	10–1000 μV	dc–1	Skin-surface electrodes
	0.5–80 mV	dc–1	Stomach-surface electrodes
Electromyography	0.1–5 mV	dc–10,000	Needle electrodes
Eye potentials			
Electro-oculogram	50–3500 μV	dc–50	Contact electrodes
Electroretinogram	0–900 μV	dc–50	Contact electrodes
Galvanic skin response	1–500 kΩ	0.01–1	Skin electrodes

Table 1.1 *(Continued)*

Parameter or Measuring Technique	Principal Measurement Range of Parameter	Signal Frequency Range, Hz	Standard Sensor or Method
Gastric pH	3–13 pH units	dc–1	pH electrode; antimony electrode
Gastrointestinal pressure	0–100 cm H_2O	dc–10	Strain-gage manometer
Gastrointestinal forces	1–50 g	dc–1	Displacement system, linear variable differential transformer
Nerve potentials	0.01–3 mV	dc–10,000	Surface or needle electrodes
Phonocardiography	Dynamic range 80 dB, threshold about 100 μPa	5–2000	Microphone
Plethysmography (volume change)	Varies with organ measured	dc–30	Displacement chamber or impedance change
Circulatory	0–30 ml	dc–30	Displacement chamber or impedance change
Respiratory functions			
Pneumotachography (flow rate)	0–600 liter/min	dc–40	Pneumotachograph head and differential pressure
Respiratory rate	2–50 breaths/min	0.1–10	Strain gage on chest, impedance, nasal thermistor
Tidal volume	50–1000 ml/breath	0.1–10	Above methods
Temperature of body	32–40 °C 90–104 °F	dc–0.1	Thermistor, thermocouple

SOURCE: Revised from *Medical Engineering*. C. D. Ray (ed.). Copyright 1974 by Year Book Medical Publishers, Inc., Chicago. Used by permission.

empirical statistical and probabilistic distribution functions. Single measurements are then compared with these *norms* (see Section 1.8).

Nearly all biomedical measurements depend either on some form of energy being applied to the living tissue or on some energy being applied as an incidental consequence of sensor operation. X-ray and ultrasonic imaging techniques and electromagnetic or Doppler ultrasonic blood flowmeters depend on externally applied energy interacting with living tissue. Safe levels of these various types of energy are difficult to establish, because many mechanisms of tissue damage are not well understood. A fetus is particularly vulnerable during the early stages of development. The heating of tissue is one effect that must be limited, because even reversible

physiological changes can affect measurements. Damage to tissue at the molecular level has been demonstrated in some instances at surprisingly low energy levels.

Operation of instruments in the medical environment imposes important additional constraints. Equipment must be reliable, easy to operate, and capable of withstanding physical abuse and exposure to corrosive chemicals. Electronic equipment must be designed to minimize electric-shock hazards (Chapter 14). The safety of patients and medical personnel must be considered in all phases of the design and testing of instruments. The Medical Device Amendments of 1976 and the Safe Medical Devices Act of 1990 amend the Federal Food, Drug, and Cosmetics Act to provide for the safety and effectiveness of medical devices intended for human use (Section 1.26).

1.5 CLASSIFICATIONS OF BIOMEDICAL INSTRUMENTS

The study of biomedical instruments can be approached from at least four viewpoints. Techniques of biomedical measurement can be grouped according to the *quantity that is sensed*, such as pressure, flow, or temperature. One advantage of this classification is that it makes different methods for measuring any quantity easy to compare.

A second classification scheme uses the *principle of transduction*, such as resistive, inductive, capacitive, ultrasonic, or electrochemical. Different applications of each principle can be used to strengthen understanding of each concept; also, new applications may be readily apparent.

Measurement techniques can be studied separately for each *organ system*, such as the cardiovascular, pulmonary, nervous, and endocrine systems. This approach isolates all important measurements for specialists who need to know only about a specific area, but it results in considerable overlap of quantities sensed and principles of transduction.

Finally, biomedical instruments can be classified according to the *clinical medicine specialties*, such as pediatrics, obstetrics, cardiology, or radiology. This approach is valuable for medical personnel who are interested in specialized instruments. Of course, certain measurements such as blood pressure are important to many different medical specialties.

1.6 INTERFERING AND MODIFYING INPUTS

Desired inputs are the measurands that the instrument is designed to isolate. *Interfering inputs* are quantities that inadvertently affect the instrument as a consequence of the principles used to acquire and process the desired inputs.

If spatial or temporal isolation of the measurand is incomplete, the interfering input can even be the same quantity as the desired input. *Modifying inputs* are undesired quantities that indirectly affect the output by altering the performance of the instrument itself. Modifying inputs can affect processing of either desired or interfering inputs. Some undesirable quantities can act as both a modifying input and an interfering input.

A typical electrocardiographic recording system, shown in Figure 1.2, illustrates these concepts. The desired input is the electrocardiographic voltage V_{ecg} that appears between the two electrodes on the body surface. One interfering input is 60 Hz noise voltage induced in the shaded loop by environmental ac magnetic fields. The desired and the interfering voltages are in series, so both components appear at the input to the differential amplifier. Also, the difference between the capacitively coupled displacement currents flowing through each electrode and the body to ground causes an interfering voltage to appear across Z_{body} between the two electrodes and two interfering voltages across Z_1 and Z_2, the electrode impedances.

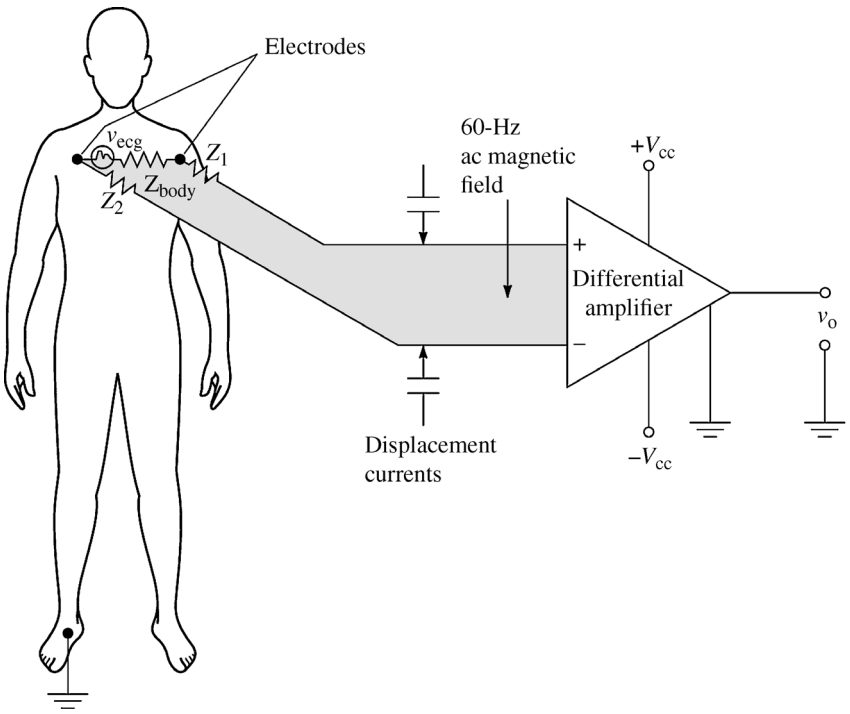


Figure 1.2 Simplified electrocardiographic recording system Two possible interfering inputs are stray magnetic fields and capacitively coupled noise. Orientation of patient cables and changes in electrode–skin impedance are two possible modifying inputs. Z_1 and Z_2 represent the electrode–skin interface impedances.

An example of a modifying input is the orientation of the patient cables. If the plane of the cables is parallel to the ac magnetic field, magnetically introduced interference is zero. If the plane of the cables is perpendicular to the ac magnetic field, magnetically introduced interference is maximal.

1.7 COMPENSATION TECHNIQUES

The effects of most interfering and modifying inputs can be reduced or eliminated either by altering the design of essential instrument components or by adding new components designed to offset the undesired inputs. The former alternative is preferred when it is feasible, because the result is usually simpler. Unfortunately, designers of instruments can only rarely eliminate the actual source of the undesired inputs. We shall discuss several compensation methods for eliminating the effects of interfering and modifying inputs.

INHERENT INSENSITIVITY

If all instrument components are inherently sensitive only to desired inputs, then interfering and modifying inputs obviously have no effect. For the electrocardiograph, an example is the twisting of the electrode wires to reduce the number of magnetic flux lines that cut the shaded loop in Figure 1.2. The voltage of the induced noise is proportional to the area of that loop. The effects of electrode motion can be reduced by techniques described in Section 5.5.

NEGATIVE FEEDBACK

When a modifying input cannot be avoided, then improved instrument performance requires a strategy that makes the output less dependent on the transfer function G_d. The negative feedback method takes a portion of the output, H_f, at any instant of time and feeds it back to the input of the instrument. This output-dependent signal is subtracted from the input, and the difference becomes the effective system input. For input, x_d, and output y, we can write

$$(x_d - H_f y)G_d = y \tag{1.1}$$

$$x_d - G_d = y(1 + H_f G_d) \tag{1.2}$$

$$y = \frac{G_d}{1 + H_f G_d} x_d \tag{1.3}$$

G_d usually includes amplification, so $H_f G_d \gg 1$ and $y \cong (1/H_f)(x_d)$. This well-known relationship shows that only the feedback element, H_f, determines the output for a given input. Of course, this strategy fails if H_f is also affected by modifying inputs. Usually the feedback device carries less power, so it is more accurate and linear. Less input-signal power is also needed for this feedback scheme, so less loading occurs. The major disadvantage of using this feedback principle is that dynamic instability leading to oscillations can occur, particularly if G_d contains time delays. The study of feedback systems is a well-developed discipline that cannot be pursued further here (Kuo and Golnaraghi, 2009).

SIGNAL FILTERING

A filter separates signals according to their frequencies. Most filters accomplish this by attenuating the part of the signal that is in one or more frequency bands. A more general definition for a filter is "a device or program that separates data, signals, or material in accordance with specified criteria" (IEEE, 1997).

Filters may be inserted at the instrument input, at some point within the instrument, or at the output of the instrument. In fact, the limitations of people's senses may be used to filter unwanted signal components coming from display devices. An example is the utilization of flicker fusion for rapidly changing images from a real-time ultrasonic scanner.

Input filtering blocks interfering and modifying inputs but does not alter the desired input. Filter elements may be distinct devices that block or pass all inputs, or they may be embodied in a single device that selectively blocks only undesired inputs. Many designers do not use input filters that are electric circuits but use instead mechanical, pneumatic, thermal, or electromagnetic principles to block out undesired environmental inputs. For example, instruments are often shock-mounted to filter vibrations that affect sensitive instrument components. Electromagnetic shielding is often used to block interfering electric and magnetic fields, such as those indicated in Figure 1.2.

Electronic filters are often incorporated at some intermediate stage within the instrument. To facilitate filtering based on differences in frequency, mixers and modulators are used to shift desired and/or undesired signals to another frequency range where filtering is more effective. Computers are used to filter signals on the basis of template-matching techniques and various time-domain signal properties. These filters may even have time- or signal-dependent criteria for isolating the desired signal.

Output filtering is possible, though it is usually more difficult because desired and undesired output signals are superimposed. The selectivity needed may be easier to achieve with higher-level output signals.

OPPOSING INPUTS

When interfering and/or modifying inputs cannot be filtered, additional interfering inputs can be used to cancel undesired output components. These extra intentional inputs may be the same as those to be canceled. In general, the unavoidable and the added opposing inputs can be quite different, as long as the two output components are equal so that cancellation results. The two outputs must cancel despite variations in all the unavoidable interfering inputs and variations in the desired inputs. The actual cancellation of undesired output components can be implemented either before or after the desired and undesired outputs are combined. Indeed, either the intentional or the unavoidable interfering input signal might be processed by G_d. The method of opposing inputs can also be used to cancel the effects of modifying inputs.

Automatic real-time corrections are implied for the method of opposing inputs just described. Output corrections are often calculated by computer methods and applied after data are collected. This requires quantitative knowledge of the interfering and/or modifying input at the time of the measurement and also of how these inputs affect the output. This method is usually cumbersome, loses real-time information, and is often used only for rather static interfering inputs, such as temperature and atmospheric pressure.

An example of using the opposing-input method is to intentionally induce a voltage from the same 60 Hz magnetic field present in Figure 1.2 to be amplified and inverted until cancellation of the 60 Hz noise in the output is achieved. An obvious disadvantage of this method is that the amplifier gain has to be adjusted whenever the geometry of the shaded loop in Figure 1.2 changes. In electronic circuits that must operate over a wide temperature range, *thermistors* (temperature-dependent resistors) are often used to counteract unavoidable temperature-dependent changes in characteristics of active circuit elements, such as transistors and integrated circuits.

1.8 BIOSTATISTICS

The application of statistics to medical data is used to design experiments and clinical studies; to summarize, explore, analyze, and present data; to draw inferences from data by estimation or hypothesis testing; to evaluate diagnostic procedures; and to assist clinical decision-making (Dawson-Saunders and Trapp, 2004).

Medical research studies can be *observational studies*, wherein characteristics of one or more groups of patients are observed and recorded, or *experimental intervention studies*, wherein the effect of a medical

procedure or treatment is investigated. The simplest observational studies are *case-series* studies that describe some characteristics of a group. These studies are without control subjects, in order only to identify questions for further research. *Case-control* observational studies use individuals selected because they have (or do not have) some outcome or disease and then look backward to find possible causes or risk factors. *Cross-sectional* observational studies analyze characteristics of patients at one particular time to determine the status of a disease or condition. *Cohort* observational studies prospectively ask whether a particular characteristic is a precursor or risk factor for an outcome or disease. Experimental clinical trials are *controlled* if the outcome for patients administered a drug, device, or procedure is compared to the outcome for patients given a placebo or another accepted treatment. The trials are *uncontrolled* if there is no such comparison. *Concurrent controls* are best, because patients are selected in the same way and for the same time period. *Double-blind* study with *randomized* selection of patients to treatment options is preferred, because this design minimizes investigator and patient bias. Medical outcome studies that show cost-effective improvements in patient health are increasingly required prior to adoption and reimbursement for new medical technologies (Anonymous, 2001).

Quantitative data are measured on a continuous or discrete *numerical scale* with some precision. Qualitative data values that fit into categories are measured on *nominal scales* that show the names of the categories. An *ordinal scale* is used when the categories exhibit an inherent order. Descriptive statistics are useful to summarize data and their attributes. *Distributions* of empirical or theoretical data reflect the values of a variable or characteristic and the frequency of occurrence of those values.

Measures of the middle, or central tendency, include the well-known *mean*, which is the sum of observed values divided by the number of observations. The mean, found as follows,

$$\overline{X} = \frac{\sum X_i}{n} \tag{1.4}$$

works best as the measure of central tendency for symmetric distributions. The *median* is the value for which half the observations are smaller and half are larger; it is used for skewed numerical data or ordinal data. The *mode* is the observation that occurs most frequently; it is used for bimodal distributions. The *geometric mean* is the *n*th root of the product of the observations:

$$\text{GM} = \sqrt[n]{X_1 X_2 X_3 \cdots X_n} \tag{1.5}$$

It is used with data on a logarithmic scale.

Measures of spread or dispersion of data describe the variation in the observations. The *range*, which is the difference between the largest and smallest observations, is used to emphasize extreme values. The *standard deviation* is a measure of the spread of data about the mean. It is computed as follows:

$$s = \sqrt{\frac{\sum (X_i - \overline{X})^2}{n-1}} \tag{1.6}$$

It is used with the mean for symmetric distributions of numerical data. Regardless of the type of symmetric distribution, at least 75% of the values always lie between $\overline{X} - 2s$ and $\overline{X} + 2s$. The *coefficient of variation* is calculated as follows:

$$\text{CV} = \left(\frac{s}{\overline{X}}\right)(100\%) \tag{1.7}$$

It standardizes the variation, making it possible to compare two numerical distributions that are measured on different scales. A *percentile* gives the percentage of a distribution that is less than or equal to the percentile number; it may be used with the median for ordinal data or skewed numerical data. The *interquartile range* is the difference between the 25th and 75th percentiles, so it describes the central 50% of a distribution with any shape. The *standard error of the mean*, SEM (standard deviation of the mean) $s_{\overline{X}} = s/\sqrt{n-1}$, expresses the variability to be expected among the *means* in future samples, whereas the *standard deviation* describes the variability to be expected among *individuals* in future samples.

EXAMPLE 1.1 Your samples from a population are 1, 1, 3, 5, 5. Estimate the mean $\overline{X}$, the standard deviation s, and the standard deviation of mean $s_{\overline{X}}$.

ANSWER Mean $\overline{X}$ = (sum of values)/(number of values) = $(1+1+3+5+5)/5 = 3$ standard deviation $s = \{[(1-3)^2 + (1-3)^2 + (3-3)^2 + (5-3)^2 + (5-3)^2]/(5-1)\}^{1/2} = (16/4)^{1/2} = 2$ standard deviation of mean $s_{\overline{X}} = s/\sqrt{n-1} = 2/\sqrt{5-1} = 1$.

Often we need to study relationships between two numerical characteristics. The *correlation coefficient* r is a measure of the relationship between numerical variables X and Y for paired observations.

$$r = \frac{\sum (X_i - \overline{X})(Y_i - \overline{Y})}{\sqrt{\sum (X_i - \overline{X})^2 (Y_i - \overline{Y})^2}} \tag{1.8}$$

The correlation coefficient ranges from –1 for a negative linear relationship to +1 for a positive linear relationship; 0 indicates that there is no linear relationship between *X* and *Y*. The correlation coefficient is independent of the units employed to measure the variables, which can be different. Like the standard deviation, the correlation coefficient is strongly influenced by outlying values. Because the correlation coefficient measures only a straight-line relationship, it may be small for a strong curvilinear relationship. Of course, a high correlation does *not* imply a cause-and-effect relationship between the variables.

Estimation and hypothesis testing are two ways to make an inference about a value in a population of subjects from a set of observations drawn from a sample of such subjects. In estimation, *confidence intervals* are calculated for a statistic such as the mean. The confidence intervals indicate that a percentage—say 95%—of such confidence intervals contain the true value of the population mean. The confidence intervals indicate the degree of confidence we can have that they contain the true mean. Hypothesis testing reveals whether the sample gives enough evidence for us to reject the *null hypothesis*, which is usually cast as a statement that expresses the opposite of what we think is true. A *P-value* is the probability of obtaining, if the null hypothesis is true, a result that is at least as extreme as the one observed. The *P-value* indicates how often the observed difference would occur by chance alone if, indeed, nothing but chance were affecting the outcome. Recent trends favor using estimation and confidence intervals rather than hypothesis testing.

Methods for measuring the accuracy of a diagnostic procedure use three pieces of information. The *sensitivity* [TP/(TP + FN)] of a test is the probability of its yielding true positive (TP) results in patients who actually have the disease. A test with high sensitivity has a low *false-negative* (FN) rate. The *specificity* [TN/(TN + FP)] of a test is the probability of its yielding negative results in patients who do not have the disease. A test with high specificity has a low *false-positive* (FP) rate; it does not give a FP result in many patients who do not have the disease. The third piece of information is the *prior probability*, or prevalence [(TP + FN)/(TN + TP + FN + FP)] of the condition prior to the test (all diseased persons divided by all persons). There are several methods for revising the probability that a patient has a condition on the basis of the results of a diagnostic test. Taking into consideration the results of a diagnostic procedure is only one part of the complex clinical decision-making process. Decision tree analysis and other forms of decision analysis that include economic implications are also used in an effort to make optimal decisions (Webster, 2004).

1.9 GENERALIZED STATIC CHARACTERISTICS

To enable purchasers to compare commercially available instruments and evaluate new instrument designs, quantitative criteria for the performance of instruments are needed. These criteria must clearly specify how well an instrument measures the desired input and how much the output depends on interfering and modifying inputs. Characteristics of instrument performance are usually subdivided into two classes on the basis of the frequency of the input signals.

Static characteristics describe the performance of instruments for dc or very low-frequency inputs. The properties of the output for a wide range of constant inputs demonstrate the quality of the measurement, including nonlinear and statistical effects. Some sensors and instruments, such as piezoelectric devices, respond only to time-varying inputs and have no static characteristics.

Dynamic characteristics require the use of differential and/or integral equations to describe the quality of the measurements. Although dynamic characteristics usually depend on static characteristics, the nonlinearities and statistical variability are usually ignored for dynamic inputs, because the differential equations become difficult to solve. Complete characteristics are approximated by the sum of static and dynamic characteristics. This necessary oversimplification is frequently responsible for differences between real and ideal instrument performance.

ACCURACY

The *accuracy* of a single measured quantity is the difference between the true value and the measured value divided by the true value. This ratio is usually expressed as a percent. Because the true value is seldom available, the accepted true value or reference value should be traceable to the National Institute of Standards and Technology.

The accuracy usually varies over the normal range of the quantity measured, usually decreases as the full-scale value of the quantity decreases on a multirange instrument, and also often varies with the frequency of desired, interfering, and modifying inputs. Accuracy is a measure of the total error without regard to the type or source of the error. The possibility that the measurement is low and that it is high are assumed to be equal. The accuracy can be expressed as percent of reading, percent of full scale, ± number of digits for digital readouts, or ±1/2 the smallest division on an analog scale. Often the accuracy is expressed as a sum of these. For example, on a digital device: ±0.01% of reading ±0.015% of full-scale ±1 digit. If accuracy is

expressed simply as a percentage, full scale is usually assumed. Some instrument manufacturers specify accuracy only for a limited period of time.

PRECISION

The *precision* of a measurement expresses the number of distinguishable alternatives from which a given result is selected. For example, a meter that displays a reading of 2.434 V is more precise than one that displays a reading of 2.43 V. High-precision measurements do not imply high accuracy, however, because precision makes no comparison to the true value.

RESOLUTION

The smallest incremental quantity that can be measured with certainty is the *resolution*. If the measured quantity starts from zero, the term *threshold* is synonymous with *resolution*. Resolution expresses the degree to which nearly equal values of a quantity can be discriminated.

REPRODUCIBILITY

The ability of an instrument to give the same output for equal inputs applied over some period of time is called *reproducibility* or *repeatability*. Reproducibility does not imply accuracy. For example, a broken digital clock with an AM or PM indicator gives very reproducible values that are accurate only once a day.

STATISTICAL CONTROL

The accuracy of an instrument is not meaningful unless all factors, such as the environment and the method of use, are considered. Statistical control ensures that random variations in measured quantities that result from all factors that influence the measurement process are tolerable. Any systematic errors or bias can be removed by calibration and correction factors, but random variations pose a more difficult problem. The measurand and/or the instrument may introduce statistical variations that make outputs unreproducible. If the cause of this variability cannot be eliminated, then statistical analysis must be used to determine the error variation. The estimate of the true value can be improved by making multiple measurements and averaging the results.

STATIC SENSITIVITY

A static calibration is performed by holding all inputs (desired, interfering, and modifying) constant except one. This one input is varied incrementally over the normal operating range, resulting in a range of incremental outputs. The static sensitivity of an instrument or system is the ratio of the incremental output quantity to the incremental input quantity. This ratio is the static component of G_d for desired inputs within the range of the incremental inputs. The incremental slope can be obtained from either the secant between two adjacent points or the tangent to one point on the calibration curve. The static sensitivity may be constant for only part of the normal operating range of the instrument, as shown in Figure 1.3(a). For input–output data that indicate a straight-line calibration curve, the slope m and intercept b for the line with the minimal sum of the squared differences between data points and the line are given by the following equations:

$$m = \frac{n\sum x_d y - (\sum x_d)(\sum y)}{n\sum x_d^2 - (\sum x_d)^2} \tag{1.9}$$

$$b = \frac{(\sum y)(\sum x_d^2) - (\sum x_d y)(\sum x_d)}{n\sum x_d^2 - (\sum x_d)^2} \tag{1.10}$$

$$y = mx_d + b \tag{1.11}$$

where n is the total number of points and each sum is for all n points. The static sensitivity for modulating sensors is usually given per volt of excitation, because the output voltage is proportional to the excitation voltage. For example, the static sensitivity for a blood pressure sensor containing a strain-gage bridge might be $50\,\mu V \cdot V^{-1} \cdot mm\,Hg^{-1}$.

ZERO DRIFT

Interfering and/or modifying inputs can affect the static calibration curve shown in Figure 1.3(a) in several ways. Zero drift has occurred when all output values increase or decrease by the same absolute amount. The slope of the sensitivity curve is unchanged, but the output-axis intercept increases or decreases as shown in Figure 1.3(b). The following factors can cause zero drift: manufacturing misalignment, variations in ambient temperature, hysteresis, vibration, shock, and sensitivity to forces from undesired directions. A change in the dc-offset voltage at the electrodes in the ECG example in Figure 1.2 is an example of zero drift. Slow changes in the dc-offset voltage do not cause a problem, because the ECG amplifier is ac-coupled. Fast changes due to motion of the subject do cause low-frequency artifact to appear at the output.

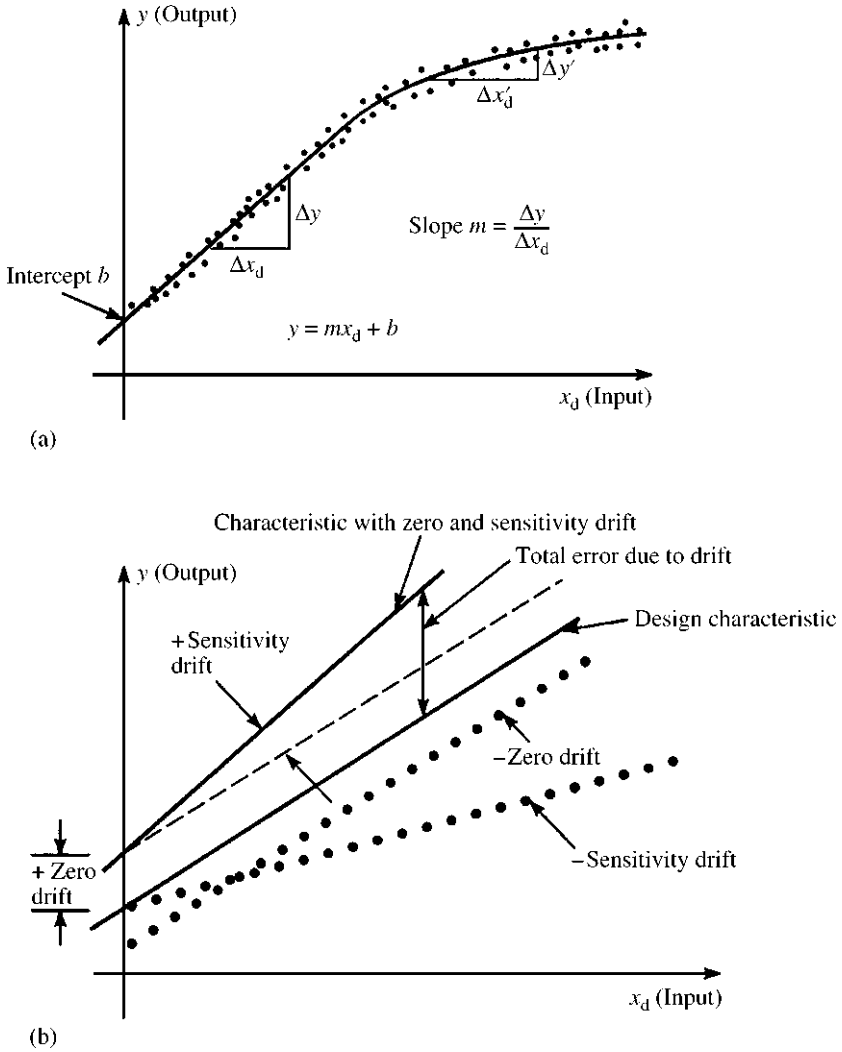


Figure 1.3 (a) Static-sensitivity curve that relates desired input x_d to output y. Static sensitivity may be constant for only a limited range of inputs. (b) Static sensitivity: zero drift and sensitivity drift. Dotted lines indicate that zero drift and sensitivity drift can be negative. [Part (b) modified from *Measurement Systems: Application and Design*, by E. O. Doebelin. Copyright 1990 by McGraw-Hill, Inc. Used with permission of McGraw-Hill Book Co.]

SENSITIVITY DRIFT

When the slope of the calibration curve changes as a result of an interfering and/or modifying input, a drift in sensitivity results. Sensitivity drift causes error that is proportional to the magnitude of the input. The slope of the calibration curve can either increase or decrease, as indicated in Figure 1.3(b). Sensitivity drift can result from manufacturing tolerances, variations in power supply, nonlinearities, and changes in ambient temperature and pressure. Variations in the ECG-amplifier voltage gain as a result of fluctuations in dc power-supply voltage or change in temperature are examples of sensitivity drift.

LINEARITY

A system or element is linear if it has properties such that, if y_1 is the response to x_1, and y_2 is the response to x_2, then $y_1 + y_2$ is the response to $x_1 + x_2$, and Ky_1 is the response to Kx_1. These two requirements for system linearity are restated in Figure 1.4(a).

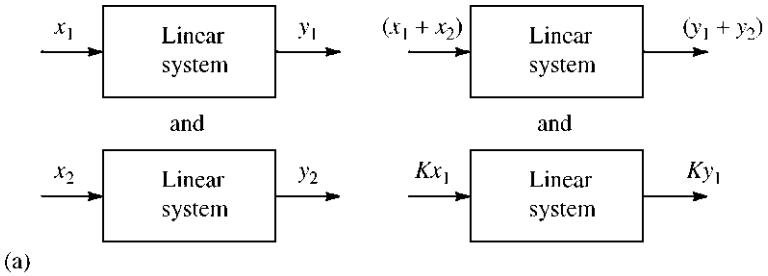
They are clearly satisfied for an instrument with a calibration curve that is a straight line.

Keep in mind, however, that high accuracy does not necessarily imply linearity. In practice, no instrument has a perfect linear response, so a measure of deviation from linearity is needed. *Independent nonlinearity* expresses the maximal deviation of points from the least-squares fitted line as either $\pm A\%$ of the reading or $\pm B\%$ of full scale, whichever is greater (that is, whichever permits the larger error). This linearity specification is shown in Figure 1.4(b). For up-scale readings, the percent-of-reading figure is desirable because most errors are proportional to the reading. For small readings near zero, however, percentage of full scale is more realistic because it is not feasible to test for small percent-of-reading deviations near zero. All data points must fall inside the "funnel" shown in Figure 1.4(b). For most instruments that are essentially linear, if other sources of error are minimal, the accuracy is equal to the nonlinearity.

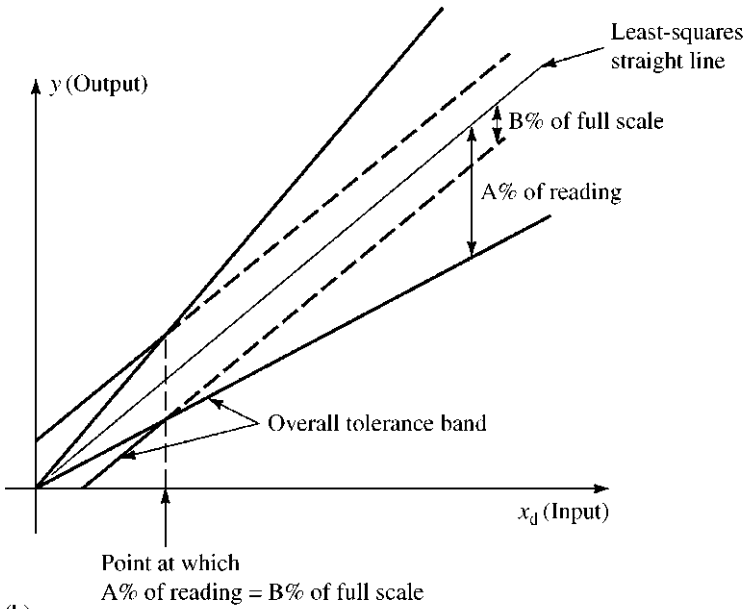
INPUT RANGES

Several maximal ranges of allowed input quantities are applicable for various conditions. Minimal resolvable inputs impose a lower bound on the quantity to be measured. The normal linear operating range specifies the maximal or near-maximal inputs that give linear outputs.

The static linear range and the dynamic linear range may be different. The maximal operating range is the largest input that does not damage the



(a)



(b)

Figure 1.4 (a) Basic definition of linearity for a system or element. The same linear system or element is shown four times for different inputs. (b) A graphical illustration of independent nonlinearity equals ±A% of the reading, or ±B% of full scale, whichever is greater (that is, whichever permits the larger error). [Part (b) modified from *Measurement Systems: Application and Design*, by E. O. Doebelin. Copyright 1990 by McGraw-Hill, Inc. Used with permission of McGraw-Hill Book Co.]

instrument. Operation in the upper part of this range is more likely to be nonlinear. Finally, storage conditions specify environmental and interfering input limits that should not be exceeded when the instrument is not being used. These ranges are not always symmetric with respect to zero input,

particularly for storage conditions. Typical operating ranges for blood pressure sensors have a positive bias, such as +200 mm Hg to −60 mm Hg (+26.6 to −8.0 kPa).

INPUT IMPEDANCE

Because biomedical sensors and instruments usually convert nonelectric quantities into voltage or current, we introduce a generalized concept of input impedance. This is necessary so that we can properly evaluate the degree to which instruments disturb the quantity being measured. For every desired input X_{d1} that we seek to measure, there is another implicit input quantity X_{d2} such that the product $X_{d1} \cdot X_{d2}$ has the dimensions of power. This product represents the instantaneous rate at which energy is transferred across the tissue–sensor interface. The generalized input impedance Z_x is the ratio of the phasor equivalent of a steady-state sinusoidal *effort* input variable (voltage, force, and pressure) to the phasor equivalent of a steady-state sinusoidal *flow* input variable (current, velocity, and flow).

$$Z_x = \frac{X_{d1}}{X_{d2}} = \frac{\text{effort variable}}{\text{flow variable}} \tag{1.12}$$

The power P is the time rate of energy transfer from the measurement medium.

$$P = X_{d1} \cdot X_{d2} = \frac{X_{d1}^2}{Z_x} = Z_x X_{d2}^2 \tag{1.13}$$

To minimize P, when measuring effort variables X_{d1}, we should make the generalized input impedance as large as possible. This is usually achieved by minimizing the flow variable. However, most instruments function by measuring minute values of the flow variable, so the flow variable cannot be reduced to zero. On the other hand, when we are measuring flow variables X_{d2}, small input impedance is needed to minimize P. The loading caused by measuring devices depends on the magnitude of the input impedance $|Z_x|$ compared with the magnitude of the source impedance $|Z_s|$ for the desired input. Unfortunately, biological source impedances are usually unknown, variable, and difficult to measure and control. Thus, the instrument designer must usually focus on maximizing the input impedance Z_x for effort-variable measurement. When the measurand is a flow variable instead of an effort variable, it is more convenient to use the admittance $Y_x = 1/Z_x$ than the impedance.

EXAMPLE 1.2 In a design, Figure E1.2 circuit consisting of a linear potentiometer to measure the arc configuration in hospital beds monitor backrest elevation which helps ensure the proper angle is maintained for patients was implemented. A 5 V excitation source was used and the length of the potentiometer was 5 cm. The measurement system used to test the arc configuration has input impedance (RL) = 1 kΩ. Assuming that the wiper is in the middle of the potentiometer whose value is 1000 Ω. What is the sensitivity of the potentiometer system? What is the error in the measurement of voltage caused due to low value of input impedance?

ANSWER Sensitivity of the potentiometer system is 1000 Ω/5 cm = 200 Ω/cm.

Applying Kirchhoff current law (KCL) we have

$$I_1 = I_2 + I_3 \tag{E1.1}$$

Applying Kirchhoff voltage law (KVL) to loop ABCA, we have

$$-500\,I_1 - 1000\,I_3 + 5 = 0 \tag{E1.2}$$

Applying KVL to loop ACA, we have

$$-500I_1 - 500I_2 + 5 = 0 \tag{E1.3}$$

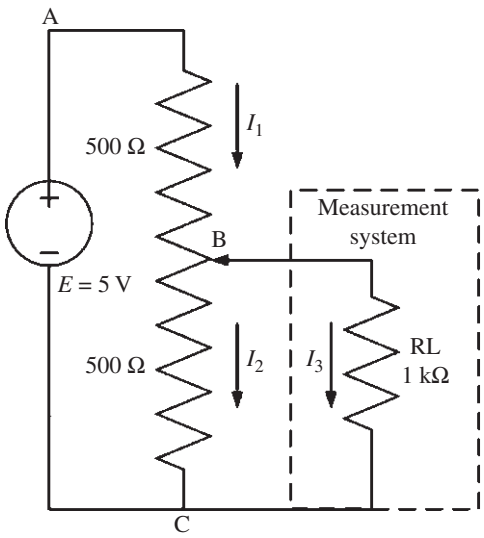


Figure E1.2 Measurement system load resistance is 1 kΩ.

Solving Eqs. (E1.1) to (E1.3), we get voltage measured will be 2 V.

$$\%\text{error} = 100(2.5 - 2)/2.5 = 20\%$$

1.10 GENERALIZED DYNAMIC CHARACTERISTICS

Only a few medical measurements, such as body temperature, are constant or slowly varying quantities. Most medical instruments must process signals that are functions of time. It is this time-varying property of medical signals that requires us to consider dynamic instrument characteristics. Differential or integral equations are required to relate dynamic inputs to dynamic outputs for continuous systems. Fortunately, many engineering instruments can be described by ordinary linear differential equations with constant coefficients. The input $x(t)$ is related to the output $y(t)$ according to the following equation:

$$a_n \frac{d_n y}{dt^n} + \cdots + a_1 \frac{dy}{dt} + a_0 y(t) = b_m \frac{d^m x}{dt^m} + \cdots + b_1 \frac{dx}{dt} + b_0 x(t) \qquad (1.14)$$

where the constants $a_i (i = 0, 1, \ldots, n)$ and $b_j (j = 0, 1, \ldots, m)$ depend on the physical and electric parameters of the system. By introducing the differential operator $D^k \equiv d^k()/dt^k$, we can write this equation as

$$(a_n D^n + \cdots + a_1 D + a_0) y(t) = (b_m D^m + \cdots + b_1 D + b_0) x(t) \qquad (1.15)$$

Readers familiar with Laplace transforms may recognize that D can be replaced by the Laplace parameter s to obtain the equation relating the transforms $Y(s)$ and $X(s)$. This is a *linear* differential equation, because the linear properties stated in Figure 1.4(a) are assumed and the coefficients a_i and b_j are not functions of time or the input $x(t)$. The equation is *ordinary*, because there is only one independent variable y. Essentially such properties mean that the instrument's methods of acquiring and analyzing the signals do not change as a function of time or the quantity of input. For example, an autoranging instrument may violate these conditions.

Most practical instruments are described by differential equations of zero, first, or second order; thus $n = 0, 1, 2$, and derivatives of the input are usually absent, so $m = 0$.

The input $x(t)$ can be classified as transient, periodic, or random. No general restrictions are placed on $x(t)$, although, for particular applications, bounds on amplitude and frequency content are usually assumed. Solutions

for the differential equation depend on the input classifications. The step function is the most common transient input for instrumentation. Sinusoids are the most common periodic function to use because, through the Fourier-series expansion, any periodic function can be approximated by a sum of sinusoids. Band-limited white noise (uniform-power spectral content) is a common random input because one can test instrument performance for all frequencies in a particular bandwidth.

TRANSFER FUNCTIONS

The transfer function for a linear instrument or system expresses the relationship between the input signal and the output signal mathematically. If the transfer function is known, the output can be predicted for any input. The *operational transfer function is the ratio* $y(D)/x(D)$ as a function of the differential operator D.

$$\frac{y(D)}{x(D)} = \frac{b_m D^m + \cdots + b_1 D + b_0}{a_n D^n + \cdots + a_1 D + a_0} \tag{1.16}$$

This form of the transfer function is particularly useful for transient inputs. For linear systems, the output for transient inputs, which occur only once and do not repeat, is usually expressed directly as a function of time, $y(t)$, which is the solution to the differential equation.

The *frequency transfer function* for a linear system is obtained by substituting $j\omega$ for D in (1.16).

$$\frac{Y(j\omega)}{X(j\omega)} = \frac{b_m (j\omega)^m + \cdots + b_1 (j\omega) + b_0}{a_n (j\omega)^n + \cdots + a_1 (j\omega) + a_0} \tag{1.17}$$

where $j = +\sqrt{-1}$ and ω is the angular frequency in radians per second. The input is usually given as $x(t) = A_x \sin(\omega t)$, and all transients are assumed to have died out. The output $y(t)$ is a sinusoid with the same frequency, but the amplitude and phase depend on ω; that is, $y(t) = B(\omega) \sin[\omega t + \varphi(\omega)]$. The frequency transfer function is a complex quantity having a magnitude that is the ratio of the magnitude of the output to the magnitude of the input and a phase angle φ that is the phase of the output $y(t)$ minus the phase of the input $x(t)$. The phase angle for most instruments is negative. We do not usually express the output of the system as $y(t)$ for each frequency, because we know that it is just a sinusoid with a particular magnitude and phase. Instead, the amplitude ratio and the phase angle are given separately as functions of frequency.

The dynamic characteristics of instruments are illustrated below by examples of zero-, first-, and second-order linear instruments for step and sinusoidal inputs.

ZERO-ORDER INSTRUMENT

The simplest nontrivial form of the differential equation results when all the a's and b's are zero except a_0 and b_0.

$$a_0 y(t) = b_0 x(t) \tag{1.18}$$

This is an algebraic equation, so

$$\frac{y(D)}{x(D)} = \frac{Y(j\omega)}{X(j\omega)} = \frac{b_0}{a_0} = K = \text{static sensitivity} \tag{1.19}$$

where the single constant K replaces the two constants a_0 and b_0. This zero-order instrument has ideal dynamic performance, because the output is proportional to the input for all frequencies and there is no amplitude or phase distortion.

A linear potentiometer is a good example of a zero-order instrument. Figure 1.5 shows that if the potentiometer has pure uniform resistance, then the output voltage $y(t)$ is directly proportional to the input displacement $x(t)$, with no time delay for any frequency of input. In practice, at high frequencies, some parasitic capacitance and inductance might cause slight distortion. Also, low-resistance circuits connected to the output can load this simple zero-order instrument. Example 1.2 shows a zero-order system and the loading effect.

FIRST-ORDER INSTRUMENT

If the instrument contains a single energy-storage element, then a first-order derivative of $y(t)$ is required in the differential equation.

$$a_1 \frac{dy(t)}{dt} + a_0 y(t) = b_0 x(t) \tag{1.20}$$

This equation can be written in terms of s as

$$(\tau D + 1) y(t) = K x(t) \tag{1.21}$$

where $K = b_0/a_0$ = static sensitivity and $\tau = a_1/a_0$ = time constant.

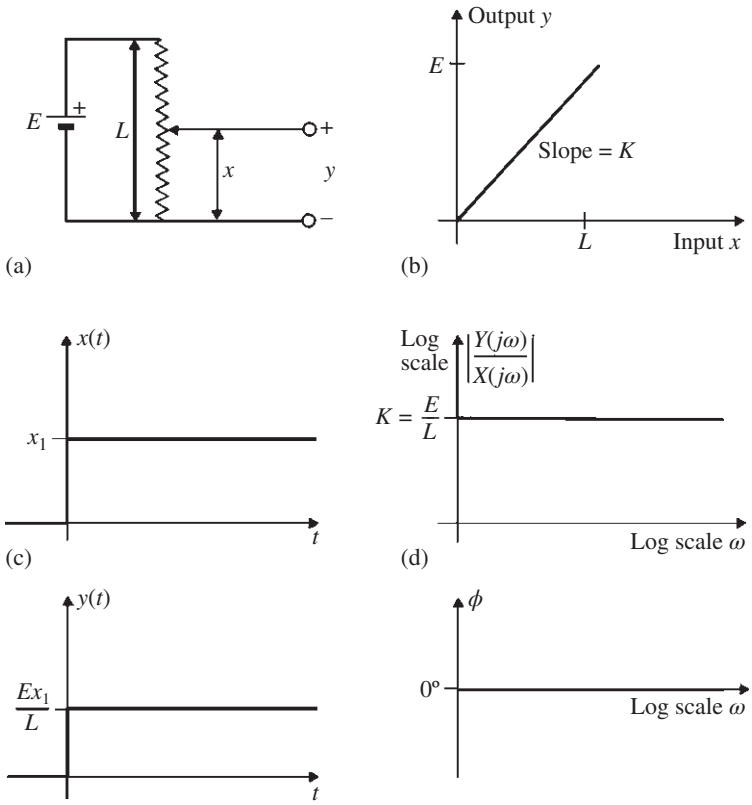


Figure 1.5 (a) A linear potentiometer, an example of a zero-order system. (b) Linear static characteristic for this system. (c) Step response is proportional to input. (d) Sinusoidal frequency response is constant with zero phase shift.

Exponential functions offer solutions to this equation when appropriate constants are chosen. The operational transfer function is

$$\frac{y(D)}{x(D)} = \frac{K}{1 + \tau D} \tag{1.22}$$

and the frequency transfer function is

$$\frac{Y(j\omega)}{X(j\omega)} = \frac{K}{1 + j\omega\tau} = \frac{K}{\sqrt{1 + \omega^2\tau^2}} \angle\phi;\ \angle\phi = \arctan(-\omega\tau/1) \tag{1.23}$$

The *RC* low-pass filter circuit shown in Figure 1.6(a) is an example of a first-order instrument. The input is the voltage $x(t)$ and the output is the voltage $y(t)$ across the capacitor. The first-order differential equation for this circuit is $RC[dy(t)/dt] + y(t) = Kx(t)$. The static-sensitivity curve given in Figure 1.6(b) shows that static outputs are equal to static inputs. This is verified by the differential equation because, for static conditions, $dy/dt = 0$. The step response in Figure 1.6(c) is exponential with a time constant $\tau = RC$.

$$y(t) = K\left(1 - e^{-t/r}\right) \tag{1.24}$$

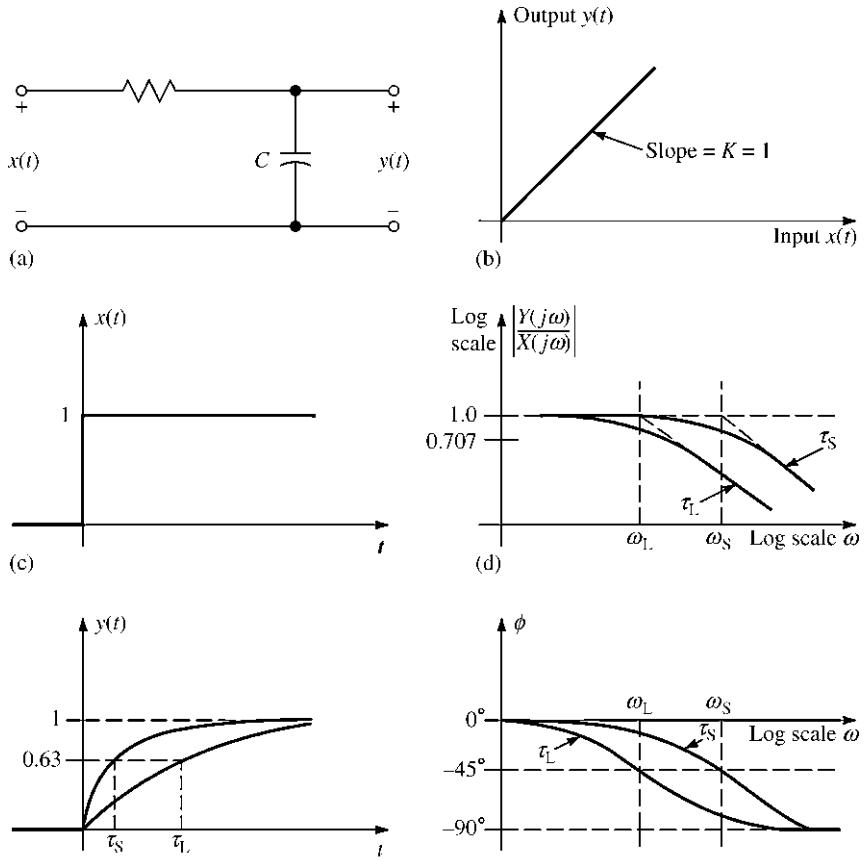


Figure 1.6 (a) A low-pass *RC* filter, an example of a first-order instrument. (b) Static sensitivity for constant inputs. (c) Step response for large time constants (τ_L) and small time constants (τ_S). (d) Sinusoidal frequency response for large and small time constants.

The smaller the time constant, the faster the output approaches the input. For sinusoids, (1.23) and Figure 1.6(d) show that the magnitude of the output decreases as frequency increases. For larger time constants, this decrease occurs at lower frequency.

When $\omega = 1/\tau$, the magnitude is $1/\sqrt{2} = 0.707$ times smaller, and the phase angle is -45°. This particular frequency ω is known as the *corner, cut-off*, or *break* frequency. Figure 1.6(d) verifies that this is a low-pass filter; low-frequency sinusoids are not severely attenuated, whereas high-frequency sinusoids produce very little output voltage. The ordinate of the frequency-response magnitude in Figure 1.6(d) is usually plotted on a log scale and may be given in units of decibels (dB), which are defined as $\text{dB} = 20\log_{10}|Y(j\omega)/X(j\omega)|$. A mercury-in-glass thermometer is another example of a low-pass first-order instrument.

EXAMPLE 1.3 A first-order low-pass instrument has a time constant of 20 ms. Find the maximal sinusoidal input frequency that will keep output error due to frequency response less than 5%. Find the phase angle at this frequency.

ANSWER

$$\frac{Y(j\omega)}{X(j\omega)} = \frac{K}{1 + j\omega\tau}$$

$$\left|\frac{K}{1 + j\omega\tau}\right| = \frac{K}{\sqrt{1 + \omega^2\tau^2}} = 0.95\,\text{K}$$

$$\left(\omega^2\tau^2 + 1\right)(0.95)^2 = 1$$

$$\omega^2 = \frac{1 - (0.95)^2}{(0.95)^2\left(20 \times 10^{-3}\right)^2}$$

$$\omega = 16.4\ \text{rad/s}$$

$$f = \frac{\omega}{2\pi} = 2.62\ \text{Hz}$$

$$\phi = \tan^{-1}\left(\frac{-\omega\tau}{1}\right) = -18.2^\circ$$

If R and C in Figure 1.6(a) are interchanged, the circuit becomes another first-order instrument known as a *high-pass filter*. The static characteristic is zero for all values of input, and the step response jumps immediately to the step voltage but decays exponentially toward zero as time increases. Thus $y(t) = Ke^{-t/\tau}$. Low-frequency sinusoids are severely attenuated, whereas high-frequency sinusoids are little attenuated. The sinusoidal transfer function is $Y(j\omega)/X(j\omega) = j\omega\tau/(1 + j\omega\tau)$.

EXAMPLE 1.4 From a 2 kV source in series with a 20 kΩ resistor, calculate the time required to charge a 100 μF defibrillator capacitor to 1.9 kV.

ANSWER Circuit is shown in Figure 1.6(a). Use (1.24) $v_C = V - Ve^{-\frac{t}{RC}}$

$$1900\,\text{V} = 2000\text{V} - 2000\text{V} \cdot e^{-\frac{t}{(20{,}000\Omega)(100\times 10^{-6}\text{F})}}$$

$$-100\,\text{V} = -2000\text{V} \cdot e^{-\frac{t}{(20{,}000\Omega)(100\times 10^{-6}\text{F})}}$$

$$0.05 = e^{-\frac{t}{(20{,}000\Omega)(100\times 10^{-6}\text{F})}}$$

$$\ln 0.05 = -\frac{t}{2\Omega \cdot \text{F}}$$

$$t = 5.99\,\text{s}$$

EXAMPLE 1.5 Temperature-guided radiofrequency catheter ablation was attempted with the catheter tip set to 70 °C to heat the affected tissue. A type J thermocouple (sensitivity of 50 μV/°C and time constant of 2 s) was used to measure this temperature. The thermocouple could be modeled as a first-order system. Write the first-order system model for this thermocouple. Use MATLAB code to plot the response of the thermocouple when it is suddenly exposed to the catheter tip temperature of 70 °C. How much time does it take to reach a steady state reading?

ANSWER The model for the first-order system is shown in (1.20) and (1.21). The thermocouple could be modeled as:

$$2\frac{dy(t)}{dx(t)} + y(t) = 50\times 10^{-6}\,x(t)$$

The transfer function for this system could be written as:

$$G = \frac{K}{\tau s + 1}$$

$$K = 50\times 10^{-6}; \quad \tau = 2\,\text{s}$$