Appendix A

Symbols and Units

Table A.1 gives many of the physical constants used in this text. The names of general variables are provided in Table A.2, along with their units. Table A.3 lists coefficients and parameters. Some more specific parameters and constants with acronyms, are provided in Table A.4. Most variables and parameters are defined locally in the chapters, sometimes a bit differently than in these tables when there is an overlap in the use of symbols. For example, in most of Chap. 8 the flow or vascular resistance is called $R_{\text{flow}}$ (Table A.3) to avoid confusion with $R$ used for radius (Table A.2). Elsewhere where there can be no confusion it is called $R$ (and is locally defined as such).

<table>
<thead>
<tr>
<th>parameter (variable)</th>
<th>value (in SI units)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Avogadro’s number ($N_A$)</td>
<td>$6.02 \times 10^{23}$ (per mole)</td>
</tr>
<tr>
<td>Boltzmann constant ($k_B$)</td>
<td>$1.381 \times 10^{-23}$ J/K</td>
</tr>
<tr>
<td>Coulomb’s Law constant ($k = 1/4\pi\varepsilon_0$)</td>
<td>$8.99 \times 10^9$ N-m²/C²</td>
</tr>
<tr>
<td>electric permittivity ($\varepsilon_0$)</td>
<td>$8.854 \times 10^{-12}$ F/m</td>
</tr>
<tr>
<td>elementary charge ($e$)</td>
<td>$1.602 \times 10^{-19}$ C</td>
</tr>
<tr>
<td>gas constant ($R = N_A k_B$)</td>
<td>$8.315$ J/mole-K</td>
</tr>
<tr>
<td>gravitation constant ($g$)</td>
<td>$9.8$ m/s² = $32.2$ ft/s²</td>
</tr>
<tr>
<td>magnetic permeability ($\mu_0$)</td>
<td>$4\pi \times 10^{-7}$ N/A²</td>
</tr>
<tr>
<td>Planck’s constant ($h$)</td>
<td>$6.626 \times 10^{-34}$ J·s</td>
</tr>
<tr>
<td>speed of light ($c$)</td>
<td>$3.0 \times 10^8$ m/s</td>
</tr>
<tr>
<td>Stefan-Boltzmann constant ($\sigma$)</td>
<td>$5.67 \times 10^{-8}$ W/m²·K⁴</td>
</tr>
</tbody>
</table>
Table A.2. General variables and units. Also see Fig. D.1

<table>
<thead>
<tr>
<th>parameter (variable)</th>
<th>common units (Definition)</th>
</tr>
</thead>
<tbody>
<tr>
<td>acceleration ($a$, $a_{decel} = -a$)</td>
<td>$m/s^2$</td>
</tr>
<tr>
<td>angular momentum ($L$)</td>
<td>kg-m$^2$/s</td>
</tr>
<tr>
<td>area ($A$, in flow $S$)</td>
<td>$m^2$</td>
</tr>
<tr>
<td>body height ($H$ or $H_b$)</td>
<td>$m^a$</td>
</tr>
<tr>
<td>body mass ($m_b$)</td>
<td>1 kg = 1,000 g</td>
</tr>
<tr>
<td>body weight ($W_b$)</td>
<td>N$^b$</td>
</tr>
<tr>
<td>charge ($q$)</td>
<td>coulombs (C)</td>
</tr>
<tr>
<td>charge, number of elementary charges ($Z$)</td>
<td>unitless, $q = Ze$</td>
</tr>
<tr>
<td>current ($I_{elect}$, $I$)</td>
<td>amps (A), 1 A = 1 C/s</td>
</tr>
<tr>
<td>current density ($J_{elect}$, $J$)</td>
<td>A/m$^2$</td>
</tr>
<tr>
<td>density (mass) ($\rho$)</td>
<td>1,000 kg/m$^3$ = 1 g/cm$^3$</td>
</tr>
<tr>
<td>density (number) ($n$)</td>
<td>#/m$^3$, #/cm$^3$</td>
</tr>
<tr>
<td>diameter ($d$, $D$)</td>
<td>1 m = 100 cm = 3.28 ft</td>
</tr>
<tr>
<td>dipole moment ($P$)</td>
<td>1 Debye (D) = $3.336 \times 10^{-30}$ C-m</td>
</tr>
<tr>
<td>distance ($L$), height ($h$, $y$)</td>
<td>$m^a$</td>
</tr>
<tr>
<td>electric field ($E$)</td>
<td>V/m</td>
</tr>
<tr>
<td>energy ($E$)</td>
<td>J$^c$</td>
</tr>
<tr>
<td>flux (particle) ($J$)</td>
<td>#/m$^2$-s</td>
</tr>
<tr>
<td>focal length ($f$)</td>
<td>m</td>
</tr>
<tr>
<td>force ($F$, $M$ (for muscle))</td>
<td>N$^b$</td>
</tr>
<tr>
<td>frequency (in space) ($k$)</td>
<td>1/m</td>
</tr>
<tr>
<td>frequency (in time) ($f$, $F$, $\nu$)</td>
<td>Hz, cycles per second (cps)</td>
</tr>
<tr>
<td>frequency (in time) (radial) ($\omega$)</td>
<td>radians per second, $\omega = 2\pi f$</td>
</tr>
<tr>
<td>heat flow, amount ($Q$)</td>
<td>kcal/h, W$^d$</td>
</tr>
<tr>
<td>heat flow, rate ($dQ/dt$)</td>
<td>W/m$^2$</td>
</tr>
<tr>
<td>intensity (acoustic, optical) ($I$)</td>
<td>$10^{-12}$ W/m$^2$</td>
</tr>
<tr>
<td>Intensity (acoustic reference) ($I_{ref}$)</td>
<td>J$^c$</td>
</tr>
<tr>
<td>kinetic energy (KE)</td>
<td>unitless</td>
</tr>
<tr>
<td>loudness ($L_P$, $L_s$)</td>
<td>phons, (10.64); sones, (10.65)</td>
</tr>
<tr>
<td>mass ($m$)</td>
<td>1 kg = 1,000 g</td>
</tr>
<tr>
<td>magnetic field ($B$)</td>
<td>1 T (T) = $10^9$ gauss (G)</td>
</tr>
<tr>
<td>magnification ($M$)</td>
<td>unitless</td>
</tr>
<tr>
<td>mobility ($\mu$)</td>
<td>m$^2$/V-s</td>
</tr>
<tr>
<td>normal force ($N$)</td>
<td>N$^b$</td>
</tr>
<tr>
<td>osmotic pressure ($\Pi$)</td>
<td>Pa$^e$</td>
</tr>
<tr>
<td>potential energy (PE)</td>
<td>J$^c$</td>
</tr>
<tr>
<td>power ($P_{power}$ or $P$, mechanical, metabolic)</td>
<td>W$^d$</td>
</tr>
<tr>
<td>pressure ($P$)</td>
<td>Pa$^e$</td>
</tr>
<tr>
<td>radiation flux ($R$)</td>
<td>W/m$^2$</td>
</tr>
<tr>
<td>radius (radius of curvature) ($r$, $R$)</td>
<td>$m^a$</td>
</tr>
<tr>
<td>reaction force ($R$)</td>
<td>N$^b$</td>
</tr>
<tr>
<td>reflection coefficient ($R_{refl}$, $R$)</td>
<td>unitless</td>
</tr>
<tr>
<td>refractive power ($P$)</td>
<td>1/m = 1 D (Diopter)</td>
</tr>
<tr>
<td>speed, angular, rotational ($\Omega$)</td>
<td>rad/s</td>
</tr>
</tbody>
</table>

(Cont.)
### Table A.2. (Continued)

<table>
<thead>
<tr>
<th>parameter (variable)</th>
<th>common units (Definition)</th>
</tr>
</thead>
<tbody>
<tr>
<td>speed, velocity (v), flow (u, v)</td>
<td>(\text{m/s}^f)</td>
</tr>
<tr>
<td>strain (\varepsilon)</td>
<td>unitless, (\text{mm/mm})</td>
</tr>
<tr>
<td>stress (\sigma)</td>
<td>(\text{Pa}^a)</td>
</tr>
<tr>
<td>temperature (T)</td>
<td>(T(\text{K}) = T(\text{°C}) + 273\°)</td>
</tr>
<tr>
<td>tension (T)</td>
<td>(\text{N}^b) (for force, as in Chap. 5)</td>
</tr>
<tr>
<td>tension (T) (surface tension)</td>
<td>(\text{N/m} (\text{force/length} (7.4)))</td>
</tr>
<tr>
<td>torque (\tau) or moment (M)</td>
<td>(\text{N-m})</td>
</tr>
<tr>
<td>transmission coefficient (T_{\text{trans}}, T)</td>
<td>unitless</td>
</tr>
<tr>
<td>volume (V) or (V_{\text{flow}})</td>
<td>(1 \text{ L} = 1,000 \text{ mL} = 1,000 \text{ cm}^3)</td>
</tr>
<tr>
<td>volume flow rate (Q)</td>
<td>(1 \text{ L/s} = 1,000 \text{ mL/s} = 1,000 \text{ cm}^3/\text{s})</td>
</tr>
<tr>
<td>voltage, potential difference (V_{\text{elect}}, V)</td>
<td>volts (V)</td>
</tr>
<tr>
<td>wavelength (\lambda)</td>
<td>(\text{m}, 1 \text{ nm} = 10^{-9} \text{ m})</td>
</tr>
<tr>
<td>work (W)</td>
<td>(\text{J}^c)</td>
</tr>
<tr>
<td>vergence (V)</td>
<td>(1/\text{m} = 1 \text{ D} (\text{Diopter}))</td>
</tr>
</tbody>
</table>

\(a\) 1 m = 100 cm = 3.28 ft, 1 mile = 5,280 ft.

\(b\) 1 N = 10^6 dynes = 0.225 lb, (Table 2.5).

\(c\) 1 J = 0.239 cal = 0.000948 BTU, 1 kcal = 4.184 J.

\(d\) 1 W = 0.86 kcal/h = 1/746 hp = 0.00134 hp (horsepower) (Table 6.1).

\(e\) 1 Pa = 1 N/m^2, 1 MPa = 1 N/mm^2 = 7,600 mmHg = 10,300 cmH_2O = 10 bar = 9.87 atm. Table 2.6.

\(f\) 1 m/s = 3.6 km/h = 3.28 fps (feet per second) = 2.24 mph (miles per hour, 1 mile = 5,280 ft).

### Table A.3. General coefficients and parameters, and units. Also see Fig. D.1

<table>
<thead>
<tr>
<th>parameter (variable)</th>
<th>common units (definition)</th>
</tr>
</thead>
<tbody>
<tr>
<td>absorption coefficient (sound, light) (\gamma)</td>
<td>(1/\text{m})</td>
</tr>
<tr>
<td>activity factor (f)</td>
<td>unitless</td>
</tr>
<tr>
<td>admittance ((Y = 1/Z = G + iB))</td>
<td>(1 \text{ mho} = 1/\text{ohm})</td>
</tr>
<tr>
<td>area moment of inertia (I_A)</td>
<td>(\text{m}^4, (4.38))</td>
</tr>
<tr>
<td>capacitance ((C_{\text{elect}} \text{ or } C)); per unit length</td>
<td>farads (F) = C/V; F/m</td>
</tr>
<tr>
<td>capacitance per area (c)</td>
<td>(\text{F/m}^2)</td>
</tr>
<tr>
<td>compliance ((C_{\text{flow}} \text{ or } C))</td>
<td>(\text{cm}^3/\text{bar}, \text{L/mmHg})</td>
</tr>
<tr>
<td>conductance (electrical, (G))</td>
<td>(\text{siemens}, 1 \text{ S} = 1/\text{ohm})</td>
</tr>
<tr>
<td>conductance per unit area (g)</td>
<td>(1/\text{ohm-m}^2)</td>
</tr>
<tr>
<td>conductivity (\sigma)</td>
<td>(1/(\text{ohm-m}), \sigma = 1/\rho)</td>
</tr>
<tr>
<td>dashpot constant (c)</td>
<td>(\text{N-s/m})</td>
</tr>
<tr>
<td>dielectric constant (\kappa)</td>
<td>unitless</td>
</tr>
<tr>
<td>diffusion coefficient (D_{\text{diff}})</td>
<td>(\text{m}^2/\text{s}, \text{cm}^2/\text{s})</td>
</tr>
<tr>
<td>distensibility (D_{\text{flow}})</td>
<td>(1/\text{Pa}, (8.20))</td>
</tr>
<tr>
<td>drag coefficient (C_D)</td>
<td>unitless</td>
</tr>
<tr>
<td>efficiency (\epsilon)</td>
<td>(0 \leq \epsilon \leq 1)</td>
</tr>
<tr>
<td>emissivity (\epsilon)</td>
<td>(0 \leq \epsilon \leq 1, 1 \text{ for a black body})</td>
</tr>
</tbody>
</table>

\(Cont.\)
Table A.3. (Continued)

<table>
<thead>
<tr>
<th>parameter (variable)</th>
<th>common units (definition)</th>
</tr>
</thead>
<tbody>
<tr>
<td>friction coefficient (static, kinetic) ($\mu_s$, $\mu_k$)</td>
<td>unitless</td>
</tr>
<tr>
<td>heat capacity ($C$)</td>
<td>kcal/°C, 1 MJ/K = 239 kcal/K</td>
</tr>
<tr>
<td>heat transfer coefficient ($h = K/d = 1/I$)</td>
<td>W/m²·°C, kcal/m²·h⁻²°C</td>
</tr>
<tr>
<td>impedance ($Z = R + iX$)</td>
<td>ohm</td>
</tr>
<tr>
<td>index of refraction ($n$)</td>
<td>unitless</td>
</tr>
<tr>
<td>insulation ($I = 1/h = d/K$)</td>
<td>m²·°C/W, m²·h⁻²°C/kcal</td>
</tr>
<tr>
<td>lift coefficient ($\rho_{ln}$)</td>
<td>unitless</td>
</tr>
<tr>
<td>moment of inertia ($I$)</td>
<td>kg·m², (3.23), (3.24)</td>
</tr>
<tr>
<td>Poisson’s ratio ($\nu$)</td>
<td>unitless, (4.7)</td>
</tr>
<tr>
<td>radius of gyration ($\rho$)</td>
<td>m</td>
</tr>
<tr>
<td>reactance ($X$)</td>
<td>ohm</td>
</tr>
<tr>
<td>resistance (flow, vascular, $R_{flow}$ or $R$)</td>
<td>mmHg·s/cm³ᵃ</td>
</tr>
<tr>
<td>resistance (electrical, $R_{elect}$ or $R$)</td>
<td>ohm ($\Omega$)</td>
</tr>
<tr>
<td>resistance (electrical; per unit length $r$)</td>
<td>ohm/m</td>
</tr>
<tr>
<td>resistivity ($\rho$)</td>
<td>ohm·m</td>
</tr>
<tr>
<td>scattering coefficient ($\alpha_{light scattering}$)</td>
<td>1/m</td>
</tr>
<tr>
<td>skin friction coefficient ($C_{sf}$)</td>
<td>unitless</td>
</tr>
<tr>
<td>specific heat ($c$)</td>
<td>kcal/kg·°Cᵇ</td>
</tr>
<tr>
<td>specific heat ratio ($\gamma$)</td>
<td>unitless, = $c_p/c_v$</td>
</tr>
<tr>
<td>speed of sound ($v_s$)</td>
<td>m/s</td>
</tr>
<tr>
<td>spring constant ($k$)</td>
<td>N/m</td>
</tr>
<tr>
<td>stroke volume ($V_{strok}$)</td>
<td>1 L = 1,000 mL = 1,000 cm³</td>
</tr>
<tr>
<td>surface tension ($\gamma$)</td>
<td>1 N/m = 1,000 dynes/cm</td>
</tr>
<tr>
<td>susceptance ($B$)</td>
<td>mho = 1/ohm</td>
</tr>
<tr>
<td>thermal conductivity ($\kappa$)</td>
<td>W/m·K</td>
</tr>
<tr>
<td>total volume flow rate ($Q_t$)</td>
<td>1 L/s = 1,000 mL/s = 1,000 cm³/s</td>
</tr>
<tr>
<td>viscosity coefficient (dynamic, absolute) ($\eta$)</td>
<td>Pa·sᵃ</td>
</tr>
<tr>
<td>viscosity coefficient (kinematic) ($\nu = \eta/\rho$)</td>
<td>Pa·s/(kg·m⁻³)</td>
</tr>
<tr>
<td>Young’s modulus ($Y$), elastic modulus ($E$)</td>
<td>Pa, 1 MPa = 1 N/mm²</td>
</tr>
</tbody>
</table>

ᵃ1 mmHg·s/cm³ = 1 mmHg·s/mL = 1 PRU.
ᵇper mass or volume, kcal/kg·°C, 1 MJ/m³·K = 239 kcal/m³·K.
ᶜ1 Pa·s (Poiseuille, PL) = 1 (N/m²)·s = 1 kg/m·s = 10 poise (P) = 1,000 cP.

Table A.4. Acronyms, including those of parameters, and units

<table>
<thead>
<tr>
<th>parameter (variable)</th>
<th>common units</th>
</tr>
</thead>
<tbody>
<tr>
<td>adenosine triphosphate, diphosphate (ATP, ADP)</td>
<td>Fig. 6.3</td>
</tr>
<tr>
<td>basal metabolic rate (BMR)</td>
<td>kcal/hᵃ</td>
</tr>
<tr>
<td>body mass index ($BMI = m_b/H^2$), Quêtelet’s index ($Q$)</td>
<td>kg/m²</td>
</tr>
<tr>
<td>center of mass (CM)</td>
<td></td>
</tr>
<tr>
<td>chromatic aberration (CA)</td>
<td></td>
</tr>
<tr>
<td>coefficient of restitution (COR)</td>
<td>unitless; (3.97)</td>
</tr>
<tr>
<td>electrocardiogram (EKG, ECG)</td>
<td>(Cont.)</td>
</tr>
</tbody>
</table>
### Table A.4. (Continued)

<table>
<thead>
<tr>
<th>Parameter (variable)</th>
<th>Common units</th>
</tr>
</thead>
<tbody>
<tr>
<td>electron transfer system (ETS)</td>
<td></td>
</tr>
<tr>
<td>focal length, back, effective, front (BFL, EFL, FFL)</td>
<td>m</td>
</tr>
<tr>
<td>focal point, plane (first: F, F'; second: F'', F'')</td>
<td></td>
</tr>
<tr>
<td>forced expiratory volume (FEV)</td>
<td>L</td>
</tr>
<tr>
<td>functional residual capacity (FRC)</td>
<td>L</td>
</tr>
<tr>
<td>Gadd Severity Index (GSI)</td>
<td>s; (3.103)</td>
</tr>
<tr>
<td>Head Injury Criterion (HIC)</td>
<td>s; (3.105)</td>
</tr>
<tr>
<td>inspiratory, expiratory reserve volume (IRV, ERV)</td>
<td>L</td>
</tr>
<tr>
<td>intraocular pressure (IOP)</td>
<td>Pa, mmHg</td>
</tr>
<tr>
<td>left atrium, ventricle (LA, LV)</td>
<td></td>
</tr>
<tr>
<td>metabolic equivalent (MET)</td>
<td>unitless</td>
</tr>
<tr>
<td>metabolic rate (MR)</td>
<td>kcal/h</td>
</tr>
<tr>
<td>near point, far point (NP, FP)</td>
<td>m</td>
</tr>
<tr>
<td>nodal point (first: N; second: N')</td>
<td></td>
</tr>
<tr>
<td>peripheral resistance unit (PRU)</td>
<td>mmHg·s/cm³</td>
</tr>
<tr>
<td>phosphocreatine (PCr)</td>
<td></td>
</tr>
<tr>
<td>physiological cross-sectional area (PCA)</td>
<td>1 cm² = 0.155 in²</td>
</tr>
<tr>
<td>principal point, plane (first: P, P'; second: P'', P''')</td>
<td></td>
</tr>
<tr>
<td>residual volume (RV)</td>
<td>L</td>
</tr>
<tr>
<td>respiration exchange ratio (RER)</td>
<td>unitless; Table 6.2</td>
</tr>
<tr>
<td>Reynolds number (Re)</td>
<td>unitless; (7.11)</td>
</tr>
<tr>
<td>right atrium, ventricle (RA, RV)</td>
<td></td>
</tr>
<tr>
<td>specific stature (S = H/m₁³), Ponderal index</td>
<td>m/kg₁/³</td>
</tr>
<tr>
<td>spherical aberration (SA)</td>
<td></td>
</tr>
<tr>
<td>Strouhal frequency, number (St)</td>
<td>unitless; (7.47)</td>
</tr>
<tr>
<td>tidal volume (TV)</td>
<td>L</td>
</tr>
<tr>
<td>total lung capacity (TLC)</td>
<td>L</td>
</tr>
<tr>
<td>total peripheral vascular resistance (TPVR)</td>
<td>mmHg·s/cm³</td>
</tr>
<tr>
<td>transient ischemic attack (TIA)</td>
<td></td>
</tr>
<tr>
<td>ultimate bending stress (UBS)</td>
<td>Pa²</td>
</tr>
<tr>
<td>ultimate compression stress (UCS)</td>
<td>Pa²</td>
</tr>
<tr>
<td>ultimate strain, ultimate percent elongation (UPE)</td>
<td>unitless</td>
</tr>
<tr>
<td>ultimate tensile stress (UTS)</td>
<td>Pa²</td>
</tr>
<tr>
<td>visual acuity (VA)</td>
<td>unitless</td>
</tr>
<tr>
<td>vital capacity, forced vital capacity (VC, FVC)</td>
<td>L</td>
</tr>
</tbody>
</table>

\(^a\)1 kcal/h = 1.162 W, 1 W = 0.86 kcal/h = 1/746 hp = 0.00134 hp (horsepower) (Table 6.1).

\(^b\)1 MPa = 1 N/mm², 1 Pa = 1 N/m², 1 MPa = 1 N/mm² = 7.600 mmHg = 10,300 cmH₂O = 10 bar = 9.87 atm. Table 2.6.
Appendix B

Locator of Major Anatomical and Anthropometric Information

This appendix cites the figures (Table B.1) and tables (Table B.2) that describe the main features of human anatomical and anthropometric information, which are used throughout this text.

Table B.1. Figures describing human anatomy and anthropometry

<table>
<thead>
<tr>
<th>figure</th>
<th>content</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.1</td>
<td>Directions, orientations, and planes</td>
</tr>
<tr>
<td>1.2</td>
<td>Anatomy of the skeletal system</td>
</tr>
<tr>
<td>1.3</td>
<td>The knee synovial joint</td>
</tr>
<tr>
<td>1.8</td>
<td>Anterior and posterior view of several large skeletal muscles</td>
</tr>
<tr>
<td>1.9</td>
<td>Antagonistic motions allowed by synovial joints</td>
</tr>
<tr>
<td>1.10</td>
<td>More antagonistic motions allowed by synovial joints</td>
</tr>
<tr>
<td>1.14</td>
<td>Ocular muscles</td>
</tr>
<tr>
<td>1.15</td>
<td>Body segment lengths</td>
</tr>
<tr>
<td>1.16</td>
<td>Postures for opposing motions</td>
</tr>
<tr>
<td>2.7</td>
<td>Bones of the arm, anterior view</td>
</tr>
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Appendix C

Differential Equations

The same form of simple differential equations is used to model very different problems throughout this text. They are presented here along with their solutions. The solutions can be checked by substituting them in the differential equation and showing that the equation is satisfied. This appendix is not meant to serve as a primer on differential equations or their solutions.

Solutions to first- and second-order differential equations, respectively, have one and two free parameters that are satisfied by the conditions of the problem. When the independent variable is time, $t$, these conditions are called initial conditions, so for the dependent variable $q(t)$, $q$ is specified at a given time, such as at $t = 0$ for a first-order differential equation. For a second-order equation, both $q$ and $\frac{dq}{dt}$ at $t = 0$ can be given. The dependent variable $q$ can be a coordinate, such as $x$ and angle $\theta$ or something else, such as force $F$. When the independent variable is a spatial coordinate, such as $x$, these conditions are called boundary conditions.

Unless otherwise specified, $F$ and $G$ are constants.

C.1 Simple First- and Second-Order Differential Equations

In these differential equations the derivatives of the dependent variable, $q$, depend on the independent variable $t$.

First-Order, Constant Driving

A variable $q(t)$ obeying

$$\frac{dq}{dt} = F$$

(C.1)
has solution

$$q(t) = Ft + q(t = 0), \quad (C.2)$$

where $q(0) = q(t = 0)$ is the initial condition.

This type of equation is used to describe the temperature rise of the body with metabolic heating and no heat loss (6.36).

Second-Order, Constant Driving

A variable $q(t)$ obeying

$$\frac{d^2 q}{dt^2} = F \quad (C.3)$$

has solution

$$q(t) = Ft^2 + \frac{dq(0)}{dt} t + q(0), \quad (C.4)$$

where the initials conditions are $q$ and $dq/dt$ evaluated at $t = 0$.

This type of equation is used in the model of ball throwing (3.76).

Second-Order, Increasing Driving

A variable $q(t)$ obeying

$$\frac{d^2 q}{dt^2} = F + Gt \quad (C.5)$$

has solution

$$q(t) = Ft^2 + \frac{Gt^3}{6} + \frac{dq(0)}{dt} t + q(0), \quad (C.6)$$

where the initials conditions are $q$ and $dq/dt$ evaluated at $t = 0$.

This type of equation is used in the model for bending a cantilever (4.44) with the position $x$ as the independent variable.

First- and Second-Order, Increasing Driving

A variable $q(t)$ obeying

$$\frac{d(t \frac{dq}{dt})}{dt} = Gt \quad (C.7)$$
or equivalently

\[ t \frac{d^2q}{dt^2} + \frac{dq}{dt} = Gt \]  \hspace{1cm} (C.8)

has solution

\[ q(t) = q(0) + \frac{Gt^2}{4}, \]  \hspace{1cm} (C.9)

where the initial condition for \( q \) is evaluated at \( t = 0 \), and \( q \) and \( \frac{dq}{dt} \) are finite at \( t = 0 \).

This type of equation is used in determining the viscous flow in a tube (7.31).

### C.2 Exponential Decay and Drag

In these differential equations the first derivative of the dependent variable, \( q \), depends on \( q \) and in some cases on the dependent variable \( t \).

**First-Order, Proportional Drag, No Driving**

A variable \( q(t) \) obeying

\[ \frac{dq}{dt} + \frac{q}{\tau} = 0 \]  \hspace{1cm} (C.10)

decays in time \( t \) as

\[ q(t) = q(0) \exp(-t/\tau), \]  \hspace{1cm} (C.11)

where \( q(0) \) is the initial condition and \( \tau \) has units of time (s) and is called a time constant. The value of \( q \) decays exponentially in time with this characteristic time constant. This can be due to a “frictional force” or damping with a rate \( 1/\tau \).

This type of equation is used in the mechanical models of non-Hookean materials ((4.22) with the strain \( \epsilon \) as the independent variable) to describe the speed when there is Stokes-type drag that is proportional to speed (7.59) and in describing pulsatile flow in (8.94), (8.105), and Problem 8.49. This equation is equivalent to the first two terms of (C.30) describing position.

This type of equation is used in the viscoelastic mechanical models of materials with constant driving terms, including the Maxwell (4.52), Voigt (4.57), and Kelvin/standard linear (4.68) models, a model of muscles (5.9), the arterial pulse (Problem 8.49), and temperature regulation (13.18).
Equation (C.10) can also be phrased as

$$\frac{dq}{dt} + \gamma q = 0,$$  \hspace{1cm} (C.12)

where the damping constant $\gamma = 1/\tau$ is defined. The solution (C.11) becomes

$$q(t) = q(0) \exp(-\gamma t).$$  \hspace{1cm} (C.13)

**First-Order, Proportional Drag, Constant Driving**

A variation of (C.10),

$$\frac{dq}{dt} + \frac{q}{\tau} = F$$  \hspace{1cm} (C.14)

with constant term $F$, has solution:

$$q(t) = (q(0) - F\tau) \exp(-t/\tau) + F\tau.$$  \hspace{1cm} (C.15)

This type of equation is used in the viscoelastic mechanical models of materials with constant driving terms, including the Maxwell (4.52), Voigt (4.57), and Kelvin/standard linear (4.68) models, a model of muscles (5.9), the arterial pulse (Problem 8.49), and temperature regulation (13.18).

**First-Order, Proportional Drag, Increasing Driving**

A variation of (C.14) includes a driving term that varies linearly with the independent variable

$$\frac{dq}{dt} + \frac{q}{\tau} = F + Gt.$$  \hspace{1cm} (C.16)

It has solution

$$q(t) = (F\tau - G\tau^2)(1 - \exp(-t/\tau)) + q(0) \exp(-t/\tau) + Gt\tau.$$  \hspace{1cm} (C.17)

This type of equation is used in the Kelvin/standard linear viscoelastic mechanical model with a linearly increasing driving term (4.72).

**First-Order, Proportional Drag, Arbitrary Temporal Driving**

A variation of (C.12) and (C.14) includes a driving term that varies arbitrarily on the independent variable

$$\frac{dq}{dt} + \frac{q}{\tau} = F(t).$$  \hspace{1cm} (C.18)
Substituting \( q(t) = s(t) \exp(-t/\tau) \) into this gives

\[
\frac{ds}{dt} = \exp(t/\tau) F(t),
\]

(C.19)

so

\[
s(t) = s(0) + \int_0^t \exp(t'/\tau) F(t') dt',
\]

(C.20)

and

\[
q(t) = \exp(-(t/\tau)) \left( q(0) + \int_0^t \exp(t'/\tau) F(t') dt' \right).
\]

(C.21)

This type of equation is used for pulsatile flow (8.103).

**First-Order, Higher-Order Drag, No Driving**

A variable \( q(t) \) obeying

\[
\frac{dq}{dt} + Aq^n = 0
\]

(C.22)

varies as

\[
q(t) = (q(0)^{1-n} + (n-1)At)^{1/(1-n)},
\]

(C.23)

for \( n \neq 1 \), where \( q(0) \) is the initial condition. For \( n = 1 \), see (C.10) and (C.11).

For \( n = -4 \) this describes flow with resistance and compliance (8.26).

For \( n = 2 \) this describes the equation of motion for hydrodynamic drag where \( q \) is speed (7.64). Then

\[
q(t) = \frac{q(0)}{1 + Aq(0)t},
\]

(C.24)

If \( q = dp/dt \), where \( p \) would be the position for this type of drag, then

\[
p(t) = p(0) + \frac{1}{A} \ln(1 + Aq(0)t).
\]

(C.25)

**C.3 Harmonic Oscillator**

In these differential equations, the second derivative of the dependent variable, \( q \), depends on \( q \) and in some cases on the dependent variable \( t \).
Harmonic Oscillator: Undamped, Not Driven

A variable $q(t)$ obeying

$$\frac{d^2 q}{dt^2} + \omega_0^2 q = 0 \quad (C.26)$$

oscillates as

$$q(t) = A \cos(\omega_0 t + \phi), \quad (C.27)$$

where $A$ is the amplitude, $\omega_0$ is the resonant frequency of this harmonic oscillator (with units rad/s), and $\phi$ is the phase. Alternatively, this solution can be expressed as

$$q(t) = B \cos(\omega_0 t) + C \sin(\omega_0 t), \quad (C.28)$$

where $B$ and $C$ are amplitudes. The frequency, $f$, is $\omega/2\pi$, and has units of Hz (Hertz) or cps (cycles per second), and so (C.27) would be

$$q(t) = A \cos(2\pi f_0 t + \phi). \quad (C.29)$$

This type of equation is used in the models of the harmonic motion of a mass on a spring (3.7), the simple (3.14) and complex pendulums (3.26), and Euler buckling (4.86).

Harmonic Oscillator: Damped, Not Driven

Adding damping to the harmonic oscillator equation (C.26) with damping constant $\gamma = 1/\tau$ gives

$$\frac{d^2 q}{dt^2} + \gamma \frac{dq}{dt} + \omega_0^2 q = 0, \quad (C.30)$$

with solution

$$q(t) = A \exp(-\gamma t/2) \cos(\omega_0 t + \phi), \quad (C.31)$$

where $A$ is the amplitude, $\omega_0$ is the resonant frequency of this harmonic oscillator (with units rad/s), and $\phi$ is the phase. This solution is not exact, but is valid for $\omega_0 \gg \gamma$. This harmonic oscillation damps in a time $\sim 1/\gamma$, which corresponds to about $\omega_0/(2\pi\gamma)$ cycles; $\omega_0/\gamma$ is often called the quality factor $Q$ of the system, as is discussed in the Chap. 10 discussion of acoustic resonances and more generally in Appendix D.

This type of equation is used in the models of harmonic oscillators, and simple and complex pendulums.
**Harmonic Oscillator: Undamped, Driven**

The equation

\[
\frac{d^2q}{dt^2} + \omega_0^2 q = F \cos(\omega t)
\]  \hspace{1cm} (C.32)

looks like the equation of motion for a simple harmonic oscillator of frequency \(\omega_0\) (C.26) with an extra term (the last one), which drives the oscillator with a “force” that oscillates at a frequency \(\omega\); \(\omega\) can differ from the resonant frequency \(\omega_0\). The particular solution to this equation is

\[
q(t) = \frac{F}{\omega_0^2 - \omega^2} \cos(\omega t),
\]  \hspace{1cm} (C.33)

to which the solution (C.27), \(q(t) = A \cos(\omega_0 t + \phi)\) (of the homogeneous equation (C.26)) is added to set the initial conditions by the proper choice of \(A\) and \(\phi\). Without the driving term \((F = 0)\), the solution is the usual harmonic solution (C.27).

This type of equation is used in the models of pulsatile blood flow (8.51) and the general models in Appendix D (D.2).

**Harmonic Oscillator: Damped, Driven**

If \(\omega_0\) were to approach \(\omega\), the response for the undamped, driven harmonic oscillator, (C.33), would approach infinity because of this resonance. There is always some damping that adds a term \(\gamma \frac{dq}{dt}\) to (C.32) to give the new equation of motion

\[
\frac{d^2q}{dt^2} + \gamma \frac{dq}{dt} + \omega_0^2 q = F \cos(\omega t).
\]  \hspace{1cm} (C.34)

This has a particular and steady-state solution

\[
q(t) = \frac{(\omega_0^2 - \omega^2) F}{(\omega_0^2 - \omega^2)^2 + (\gamma \omega)^2} \cos(\omega t).
\]  \hspace{1cm} (C.35)

The homogeneous solution (C.31), \(q(t) = A \exp(-\gamma t/2) \cos(\omega_0 t + \phi)\) for \(\omega_0 \gg \gamma\), is added to this to set the initial conditions by the proper choice of \(A\) and \(\phi\). Without the driving term \((F = 0)\), the solution is the usual damped harmonic solution (C.31).

This type of equation is used in the models of pulsatile blood flow (8.53), acoustic impedance (10.21), and the general models in Appendix D (D.2).

**C.4 Partial Differential Equations**

Partial differential equations contain derivatives of more than one independent variable.
The Diffusion Equation

The diffusion equation (7.53) in one dimension \((x)\) has the form

\[ D \frac{\partial^2 q}{\partial x^2} = \frac{\partial q}{\partial t}. \]  

(C.36)

The formal solution gives \(q(x, t)\) from \(q(x', t = 0)\), the distribution for all \(x\) (called \(x'\)) at an earlier time (defined as \(t = 0\)). It is

\[ q(x, t) = \frac{Q}{\sqrt{4\pi Dt}} \int_{-\infty}^{\infty} q(x', 0) \exp\left(-\frac{(x - x')^2}{4Dt}\right) dx', \]  

(C.37)

where \(Q\) is the integral of \(q\) over all \(x\) at any time – which means that the total amount of the entity undergoing diffusion, such as the mass or number of particles, does not change during diffusion.

The importance of this diffusion is most simply seen when the initial distribution is gaussian and has an initial spread \(\sigma(0)\),

\[ q(x, 0) = \frac{Q}{\sqrt{2\pi\sigma^2(0)}} \exp\left(-\frac{x^2}{2\sigma^2(0)}\right). \]  

(C.38)

Then the solution becomes

\[ q(x, t) = \frac{Q}{\sqrt{2\pi\sigma^2(t)}} \exp\left(-\frac{x^2}{2\sigma^2(t)}\right), \]  

(C.39)

where

\[ \sigma^2(t) = \sigma^2(0) + 2Dt. \]  

(C.40)

If the initial spread is not gaussian, the solution is slightly different but approaches this for large \(x\) and/or large \(t\). (Sometimes \(\sigma\) is defined a bit differently than it is here, as in (7.55).)

This type of equation is used in diffusion (7.53).

The integral over a gaussian probability curve,

\[ \text{erf}(x) = \frac{2}{\pi^{1/2}} \int_{0}^{x} \exp(-z^2) dz, \]  

(C.41)

is known as the error function. It increases from 0 to 1 as \(x\) increases from 0 to \(\infty\). The error function is used in the statistics describing head injury as in Fig. 3.59.

The Poisson–Boltzmann Equation

The Poisson–Boltzmann Equation (12.44) is of the form

\[ \nabla^2 q = \kappa^2 q. \]  

(C.42)
where $\nabla^2$ is the Laplacian. In one dimension $\nabla^2 q = \partial^2 q/\partial x^2$, while in Cartesian coordinates in three dimensions it is $\nabla^2 q = \partial^2 q/\partial x^2 + \partial^2 q/\partial y^2 + \partial^2 q/\partial z^2$. In three dimensions it can be expressed as $\nabla^2 q = (1/r)(d^2 (rq)/dr^2)$ when there is no angular dependence (spherical symmetry), where $r$ is the radial coordinate. Using this in (C.42) gives

$$\frac{1}{r} \frac{d^2 (rq)}{dr^2} = \kappa^2 q.$$  \hspace{1cm} \text{(C.43)}$$

Replacing $rq(r)$ by $p(r)$, this reduces to

$$\frac{d^2 p}{dr^2} = \kappa^2 p.$$  \hspace{1cm} \text{(C.44)}$$

with solution $p(r) = p(0) \exp(-\kappa r)$ valid for all $r$, and so

$$q(r) = q(0) \frac{\exp(-\kappa r)}{r}.$$  \hspace{1cm} \text{(C.45)}$$

This is used to determine the potential of a charge in a neutral region with mobile charges, as in (12.44) and Problem 12.8.
Appendix D

Similar Model Systems

This appendix describes the models used throughout the text to describe mechanical, fluid flow, electrical, and acoustic systems. Figure D.1 shows the analog in the driving forces, currents, resistances, capacitance, and inductance in each of these models.

There are many examples of these models in the text. Chapter 4 covers the spring model of the elastic properties of materials (Fig. 4.3), the dashpot model of the viscous properties of materials (Fig. 4.48), the viscoelastic model of mechanical properties of materials, including the Maxwell (Fig. 4.52), Voigt (Fig. 4.57), and Kelvin/standard linear (Fig. 4.68) models. In Chap. 5, a mechanical model of muscles, with springs and dashpots (Fig. 5.9) is presented. Models of fluid flow are described in Chaps. 7 and 8. In fact, the Windkessel models of circulation in Chap. 8 are explicitly expressed in terms of electrical components (Fig. 8.57). The mechanical and flow model of breathing in Fig. 9.16b includes compliance, resistance, and inertance. Acoustic impedance (Fig. 10.19) and admittance (Fig. 10.20) are described in Chap. 10 (and in the problems in that chapter) in relation to mechanical analogs. In Chap. 10 there also are mechanical models of vibrations in the vocal tract (as vibrations in pipes and voice filtering theory) and vocal folds (vibrations in strings, (10.41), and mechanical model with mass, springs, and dashpots (Fig. 10.13) and the two-tube models of vowel formation (Fig. 10.25). In that chapter there is also a mechanical model of the outer and inner ears (Fig. 10.28), and the vibrations of the eardrum (Fig. 10.56), tapered, uncoiled cochlea (Fig. 10.38), and hair cells (Fig. 10.57). Axon nerve conduction in Chap. 12 involved a distributed model. (See Sect. D.1)

Equivalent mechanical, electrical, and acoustic models are shown in Fig. D.2 of a typical system. The electrical model is described by

\[ V = L \frac{dI}{dt} + IR + \frac{q}{C}, \]  

(D.1)

for a voltage \( V \) producing a current \( I \); \( I = \frac{dq}{dt} \) where \( q \) is the charge. (The subscripts specific for the electrical model are omitted for simplicity.)
Fig. D.1. Model symbols are shown for (a) mechanical, (b) fluid flow, (c) electrical, and (d) acoustic models, along with the parameters and common units for each. The mechanical model is for linear (rectilinear) motion. Analogous parameters exist for the rotational mechanical model, such as for a pendulum. Viscosity is also important in mechanical models. Also see Table D.1 below. (Based on [609] and [610])

The other models are described similarly, simply by changing the parameters. Equation D.1 also be written as the second-order differential equation

\[ V = L \frac{d^2q}{dt^2} + R \frac{dq}{dt} + \frac{q}{C}, \tag{D.2} \]

as in (C.30) and (C.34).

The general solution for an oscillating voltage \( V(t) = V_0 \exp(i\omega t) \) is

\[ I(t) = \frac{V_0 \exp(i\omega t)}{R + i\omega L + 1/(i\omega C)} = \frac{V(t)}{Z}, \tag{D.3} \]

Fig. D.2. Equivalent (a) mechanical, (b) electrical, and (c) acoustic models. (Based on [609] and [610])
Appendix D Similar Model Systems

Fig. D.3. Resonant response

with complex impedance

\[ Z = R + i\omega L + \frac{1}{i\omega C}. \]  (D.4)

The resonant frequency is seen in Fig. D.3 given by

\[ \omega_{\text{res}} = \frac{1}{\sqrt{LC}} \]  (D.5)

in rad/s and \( f_{\text{res}} = 1/(2\pi\sqrt{LC}) \) in Hz or cps, and the quality factor \( Q \) is given

\[ Q = \frac{\omega_{\text{res}}}{L} \]  (D.6)

Using the notation of (C.30) and (C.34)

\[ Q = \frac{\omega_{\text{res}}}{\gamma}. \]  (D.7)

The full width of the resonance (between the points at half-maximum response) is \( \omega_{\text{res}}/Q \), as is illustrated in Fig. D.3. (This is actually the full width only for sharp resonances, for which \( \omega_{\text{res}} \gg \gamma \) and \( Q \gg 1 \).) This full width is sometimes called the bandwidth (when expressed as \( f \) in Hz), \( \Delta f \), and so an alternative definition of \( Q \) is \( Q = f/\Delta f \); this is an equivalent definition in the low-loss, high \( Q \) limit.

After the excitation is turned off ((C.34) becoming (C.30)), the energy in the system exponentially decays to \( 1/e \) of the initial value in a time \( t = 1/\gamma = Q/\omega_{\text{res}} \), where \( \gamma \) is the damping rate (as in (C.30)), which is \( R/L \) is here. This decay occurs in \( Q/2\pi \) oscillation periods. This is consistent with the definition of \( Q \) as \( 2\pi \) (energy stored)/(energy dissipated per cycle).
Appendix D Similar Model Systems

Table D.1. Analog of blood flow and electrical circuits (with units)

<table>
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<tr>
<th>Blood circulation parameter</th>
<th>Electrical parameter</th>
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</tr>
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<tbody>
<tr>
<td>volume, $V_{flow}$ (m$^3$)</td>
<td>charge, $q$ (C, coulomb)</td>
<td></td>
</tr>
<tr>
<td>blood flow rate, $Q$ (m$^3$/s)</td>
<td>current, $I$ (A, ampere)</td>
<td></td>
</tr>
<tr>
<td>pressure, $\Delta P$ (N/m$^2$)</td>
<td>voltage, $V_{elect}$ (V, volt)</td>
<td></td>
</tr>
<tr>
<td>vascular resistance, $R_{flow}$ (N-s/m$^3$)</td>
<td>resistance, $R_{elect}$ (Ω, ohm)</td>
<td></td>
</tr>
<tr>
<td>inductance, $L_{flow}$ (kg/m$^4$)</td>
<td>inductance, $L_{elect}$ (H, henry)</td>
<td></td>
</tr>
<tr>
<td>compliance, $C_{flow}$ (m$^5$/N-s)</td>
<td>capacitance, $C_{elect}$ (F, farad)</td>
<td></td>
</tr>
</tbody>
</table>

D.1 Distributed vs. Lumped Models: Electrical Analogs of Blood Flow (Advanced Topic)

So far we have discussed lumped parameter models in this appendix. In (8.2) and (8.11) flow was analyzed with the vessel as a “lumped” parameter. We have also examined cases in this text in which the parameters are distributed per unit length, such as flow resistance per unit length for volumetric flow along an artery in Chap. 8 ((8.14) and (8.25)) and electrical resistance per unit length for current flow along an axon in Chap. 12 ((12.60) and (12.67)). These are “distributed” or “transmission-line” models.

A discretized version of the distributed electrical model is shown in Fig. 12.17. Let us say that each repeated section has (very short) length $\Delta x$. The changes in electrical voltage (the driving force) and current (the response) (Table D.1) along this length of an electrical cable are described by

$$V_{elect}(x + \Delta x) - V_{elect}(x) = \frac{\partial V_{elect}}{\partial x} \Delta x = L_{elect} \frac{\partial I}{\partial t} + I R_{elect} \tag{D.8}$$

$$I(x + \Delta x) - I(x) = \frac{\partial I}{\partial x} \Delta x = C_{elect} \frac{\partial V_{elect}}{\partial t} + \frac{V_{elect}}{R_{elect}} \tag{D.9}$$

$V_{elect}$ and $I$ are functions of $x$ and $t$. The resistance, inductance, and capacitance are those for this length $\Delta x$, and can also vary with $x$. These equations can be obtained using Kirchhoff’s Laws (the 2nd and 1st laws, respectively). They were derived and then combined in the discussion of electrical signals along nerves in Chapter 12 (Fig. 12.17) to give the telegraph equations. Part of the first equation is Ohm’s Law: $\Delta V_{elect} = (\partial V_{elect}/\partial x) \Delta x = I R_{elect}$.

The analogous equations for blood flow along a vessel of length $\Delta x$ are:

$$P(x + \Delta x) - P(x) = \frac{\partial P}{\partial x} \Delta x = L_{flow} \frac{\partial Q}{\partial t} + Q R_{flow} \tag{D.10}$$

$$Q(x + \Delta x) - Q(x) = \frac{\partial Q}{\partial x} \Delta x = C_{flow} \frac{\partial P}{\partial t} + \frac{P}{R_{flow}} \tag{D.11}$$
where now the pressure is the driving force for the blood flow rate $Q$ and $R_{\text{flow}}$ is the vascular resistance. Without the inertance term, the first equation is just Poiseuille’s Law (7.25): $\Delta P = (\partial P/\partial x)\Delta x = Q R_{\text{elect}}$. The flow parameters are

$$R_{\text{flow}} = \frac{8\pi \eta L}{A^2} \quad \text{(D.12)}$$
$$C_{\text{flow}} = \frac{3LA(1 + r/w)^2}{Y(1 + 2r/w)} \quad \text{(D.13)}$$
$$L_{\text{flow}} = \frac{\rho L}{A} \quad \text{(D.14)}$$

where $A$ is the cross-sectional area of the vessel, $L$ is its length (which is $\Delta x$ for the discretized model), $r$ is its radius, $w$ is its wall thickness, and $\rho$ is the blood mass density. These equations are useful for tracking blood flow within vessels with both resistive and compliant properties.

All models of materials (and systems and processes) can be improved mathematically by adding more terms, such as in the mechanical model in Fig. 5.13. The bigger issues are whether the elements in such simple or more complex models correspond to the physical components of the material. Even if they do not, it is still important to learn if the model can be used to predict operation correctly when conditions are changed.
Appendix E

Biophysics of the Human Body

This appendix places the contents of this text within the field of biophysics.

Biophysics is hard to define well, as is illustrated by the many definitive, yet different definitions of biophysics provided in [611, 612, 613, 614, 615, 616, 617, 618, 619, 620, 621, 622]. Broadly speaking, biophysics is the applications of physics and physical principles to biology. In this context, virtually everything presented in this book is biophysics. However, this term is often used in the more restricted sense of the use of physics at a more molecular and cellular level. We will use this narrower context for the rest of this appendix, and in this restricted sense many topics covered here are still biophysics, but many areas in biophysics have not been covered. Yet another definition of biophysics is the study of biology using physical methods. This is distinguished from biological physics, which is the study of the physical properties of biology.

One topic in biophysics is the molecular structure of biological systems. This includes the electrostatics of ions in solutions (Chap. 12), the structure of biomacromolecules, such as proteins and areas such as protein folding, structure, and properties of interfaces between biological media such as cell membranes (surface tension in Chaps. 7 and 9, nerve cell membranes in Chap. 12), and ion channels in membranes (which is very briefly touched in nerve conduction in Chap. 12).

Statistical mechanics is the examination of systems composed of many similar objects or systems, each of which is well characterized. The whole ensemble of systems often is in thermal equilibrium and, after statistical mechanical analysis, can be treated by using thermodynamics. The treatment of ions in solution in Chap. 12 is the result of statistical mechanics. As stated in Chap. 5, the Hill force–velocity curve of muscles can be derived from statistical mechanical analysis of the many actin–myosin cross bridges (which was not done here). Many aspects of cell membranes and protein structure – such as folding – can be examined by using statistical mechanical methods.

Biophysics includes the bioenergetics of the photosynthesis process and the synthesis of ATP and its use. In Chap. 6 we examined the biophysics of energy usage in the human body. The movement of organisms, such as bacteria
motion and muscular movement, are part of molecular and cellular biophysics; the microscopic basis of muscle operation was explored in Chap. 5. The electrochemical properties of cell membranes and nerve signals, as discussed in Chap. 12, have always been central topics in biophysics. Some include within biophysics the higher-level integration and combinations of molecular and cellular systems, such as memory, control of movement, visual integration, and consciousness and thinking.

The use of physical characterization to biological problems plays a central role in biophysics, such as the use of X-ray diffraction (XRD) to determine molecular structure, nuclear magnetic resonance (NMR) to study molecules in more natural environments than X-ray diffraction can be used, scanning tunneling microscopy (STM) to examine the atomic structure of surfaces, atomic force microscopy (AFM) to examine surfaces and to measure forces, and optical tweezers to manipulate molecules. Both AFM and optical tweezers have been instrumental in studying the fundamental interactions in muscles, such as individual actin–myosin cross bridges (Chap. 5).
Solutions to Selected Problems

Problems of Chapter 1

1.3 Medial.

1.13 Head.

1.25 (a) (partial answer) For lower legs 3.72–9.30 kg and 5.53–9.55 kg.

1.30 (b) 0.25 m and 0.50 m.

1.31 (partial answer) Surface area is 20.1 sq ft.

1.44 (a) 92.6 kg (204.2 lb) for Man A and 84.3 kg (185.9 lb) for Man B; (b) 0.4 kg (0.9 lb) for Man A and 8.7 kg (19.2 lb) for Man B.

1.49 (b) 21.1 and 27.0.

1.57 Bigger in cold climate.

Problems of Chapter 2

2.1 Third class lever.

2.2 Triceps brachii, second class lever.

2.3 (a) First; (b) second; (c) third class levers.

2.11 (b) $m_{\text{leg}}(x_{\text{extended leg}} + x_{\text{balancing leg}}) = (m_{\text{torso+head}} + 2m_{\text{arm}})x_{\text{upper body}}$.

2.13 (a) $-m_b x$.

2.14 $T_1 = W_1 = 223.6$ N, $T_2 = W_2 = 282.8$ N, $\alpha = 26.6^\circ$.

2.17 (b) 310.6 N.

2.21 (b) $T = 131$ lb, $F = 208$ lb, angle of F is $29.8^\circ$. 
2.33 (partial answer) \(-6\) N-m for 20 cm deep.
2.34 (partial answer) \(-6.9\) N-m for upright, \(-19.25\) N-m for bent.
2.35 (partial answer) \(-21.25\) N-m for bent over, with bent knees and the object far from her body.

Problems of Chapter 3

3.6 0.27, easy to achieve with cleated running shoes.
3.10 1.7127 kg-m$^2$.
3.26 2/3.
3.35 Yes, by \(-0.07\) m, yes.
3.36 For (a) 0.254$H$; (b) 0.150$H$; (c) 0.077$H$; (d) 0.150$H$; (e) 0.254$H$; 14 cm.
3.40 Yes, because how fast the body can take off at a given angle depends on the construction of the body’s feet and legs.
3.64 44.0°, 31.1 m/s.
3.72 (partial answer) Elastic collisions are likely fatal for collision times <120 ms.
3.78 (a) 150 ft/s$^2$, 4.7$g$; (b) 1,150 lb.
3.80 1.44, 1.06, 1.01, and 0.18 m.
3.81 0.50, 0.53, and 0.55.
3.88 48.0 mph before, 24.7 mph after.
3.89 40.5 oz and about 80 mph.

Problems of Chapter 4

4.2 (c) 480 MPa.
4.5 (a) 30 Pa, (b) 67 mm$^2$, (c) 1%.
4.6 (partial answer) 1.6 MPa for nails.
4.7 (partial answer) 0.0031 for nails.
4.8 (partial answer) 8,000 N/m$^3$ for nails.
4.9 (partial answer) 780 N/m$^3$ for nails.
4.15 (a) Tension.
4.20 (a) $\lambda = 2$, $\epsilon_{\text{small}} = 1$, $\epsilon_{\text{general}} = 3/2$. 
4.21 (a) $\lambda = 2$, $\epsilon_{\text{small}} = 1$, $e = 3/8$.

4.22 (a) $\lambda = 2$, $\epsilon_{\text{small}} = 1$, $\ln \lambda = 0.69$.

4.23 (a) Yes, because the dashpot resistive force increases with speed;
(b) Yes, because the spring supplies the needed restoring force to return it to its equilibrium position.

4.24 Length is 3.1 cm, $dx/dt$ is 0.05 cm/s.

4.28 (a) $2\theta(t + 1) - 2\theta(t - 3)$, with all times in seconds.

4.33 It becomes the Voigt model, the Maxwell model, and a dashpot, respectively.

4.42 (a) 6,900 N.

Problems of Chapter 5

5.8 (a) 0.11, it is larger than the 0.09 listed in the table—but in linear theory it would be expected to be $UTS/Y = 0.22$, (b) $f_Y = 135$ N/cm$^2$, (c) the diameter of the tendons is 0.073 times that of the muscle, (d) 2,800 N/cm$^2 = 28$ MPa, which is less than the 54 MPa UTS listed in Table 4.2, so it is less than it, even with linear theory.

5.10 (a) (partial answer) 2,770 W.

5.15 The muscles are fairly near their optimal lengths. However, there are significant changes in the lengths during bicycling, but less than the maximum expected for muscles are shown for several reasons: (a) The decreases in contracted muscle length are actually greater than those shown because tendon extension will lessen the decrease in the (plotted) total muscle/tendon length, (b) the bicycle is set to use muscles in their optimal state, both in muscle length and speed, so the muscles will not be much longer or shorter than their optimal length.

5.18 (a) (partial answer) $\sqrt{3}NF_{\text{fiber}}/2$.

5.23 (a) 40 s.

Problems of Chapter 6

6.1 (partial answer) 0.3°C.

6.4 (b) 144 BTU.

6.6 (a) (partial answer) 7.1 kcal/g.

6.14 (a) 261 g, compared to 260 g and 280 g; (b) 45%; (c) 18%.
6.17 90% of the fruit and 40% of the dried fruit is water plus non-metabolizable matter.

6.24 2,200 kcal.

6.36 (a) 139 kg; (b) This is much more than the body mass of 60 kg; (c) 1,400 cycles/day, 0.95 cycles/min.

6.30 (a) −12.6 kcal/mol; (b) −14.0 kcal/mol.

6.31 66–70%.

6.41 147 moles of ATP.

6.47 6.9.

6.56 (partial answer) 580 kcal/h for 50 kg college-age women, using 40 mL/kg-min and a calorific equivalent of 4.83 kcal/L O₂.

6.57 (a) 89.9 m/min, 1.50 m/s, 3.35 mph, very good agreement; (b) 259 kcal/hr.

6.61 (partial answer) (a) 1048 kcal/h and 3.6 L O₂/min for Stage VIII, (b) 35 kcal for Stage VIII.

6.64 (a) 305 J, (b) 36 kcal, (c) no, but it excludes the pushing for 5–10 s during each play, which obviously accounts for most of the energy expenditure (although relatively little of the work done).

6.67 12.2 kJ, 11.7 kcal.

6.74 Activity factor is 1.48 (assuming “self-care” walking and also cycling at 5.5 mph), MR is 1920 kcal/day (using 1300 kcal/day BMR).

6.75 (a) 4.9, (b) 4.2.

6.80 15 kg, 33 lb.

6.88 2.2 L/h.

6.95 (a) 70 kcal/h.

6.100 \[ T_{we} = 35.74 + 0.6215T - 35.75u^{0.16} + 0.4275T u^{0.16}. \]

6.101 (a) A and B; (b) C; (c) A.

6.106 The first term is 40 kcal/day, second term in 220 kcal/day. The second term is very significant.

Problems of Chapter 7

7.3 2,240 cm³, 2.49 kg, 24.4 N; using a mass density of 1.11 g/cm³.

7.7 No, his density is then 1.01 g/cm³, which is above that of water, using a fat density of 0.8 g/cm³.
7.18 (a) $Q/N$; (b) The diameter of the small tubes would then be $N^{1/4}D$, which is not possible because the diameter of the small tubes would then exceed that of the larger tubes.

7.25 (partial answer) 0.32 cm in a gas.

7.38 $Re = 0.001$, viscous/laminar.

Problems of Chapter 8

8.5 You should be concerned, but not about your blood pressure (which is really $120 \text{mmHg}/80 \text{mmHg}$), but about the person who told you your blood pressure in absolute pressure instead of the standard gauge pressure.

8.12 (a) 121 mmHg, which is 32% higher than the 92 mmHg base case, (b) 115 mmHg, which is 25% higher.

8.15 (a),(b) They change by a factor of 1/8.

8.19 $1 \text{N-s/m}^5 = 10^6 (\text{N/m}^2)/(\text{cm}^3/\text{s}) = 10^5 \text{dyne-s/cm}^5 = 1.32 \times 10^8 \text{PRU}.$

8.27 $u_1/4$.

8.34 98.6 mmHg, using a blood density of 1060 kg/m$^3$.

8.39 Type I skeletal muscle, because very fast response is not needed and endurance is essential.

8.43 0.128 L = 128 cm$^3$.

8.45 (partial answer) 3.6 cm inner radius.

8.53 Heart beat rate: 83/min for the man, 91/min for the woman, 161/min for the infant.

Problems of Chapter 9

9.6 (partial answer) 4,720 for $z = 0$ (turbulent), 0.18 for $z = 20$ (laminar).

9.16 0.0078 cmH$_2$O/(L/s), which is much smaller than the total resistance of $\sim 2$ cmH$_2$O/(L/s).

9.20 2.7 cmH$_2$O-s/L.

9.26 Larger in a mouse (0.005 L/kg-cm-H$_2$O) than in man (0.003 L/kg-cm-H$_2$O).

9.33 (partial answer) 570 L of O$_2$ (at 1 atmosphere oxygen pressure) are consumed per day.

9.37 (partial answer) 225 mmHg total, 42 mmHg oxygen.
Problems of Chapter 10

10.4 Respectively at 0, 20, and 25°C, \( \nu_{\text{air}} \) is 331, 343, and 346 m/s; \( \rho_{\text{air}} \) is 1.292, 1.204, and 1.184 kg/m\(^3\); and \( Z_{\text{air}} \) is 428, 413, and 410 kg/m\(^2\)-s.

10.6 0.00002–2000 dyne/cm\(^2\).

10.11 70 dB SPL.

10.14 (a) 100 dB SPL, 40 dB SPL.

10.20 (a) 0.27.

10.33 \( \sim 1,000 \) Hz, which makes sense since the voices of children are higher pitched than those of adults.

10.38 160 Hz, assuming a mass with the mass of the vocal folds is attached to a massless spring with the force constant of the vocal folds. This really requires the analysis of a freely oscillating massive spring, which shows that oscillation frequency is \( \pi/2 \times \) this value [458].

10.40 Lower, because they have higher fundamental buzzing frequencies.

10.47 \( 3 \times 10^{-13} \) m.

10.49 It is \( 9 \times 10^6 \) larger at 3,000 Hz than that at 1 Hz.

10.55 (a) Radii of 0.4 \( \mu \)m for 20 Hz to 400 \( \mu \)m for 20 kHz.

10.60 17,000 Hz would be best because the auditory sensitivity of older people is very low at this frequency relative to that of younger people, at 250 Hz and 1,000 Hz the auditory sensitivity is not that different for older and younger people, at 30,000 Hz humans have no auditory sensitivity.

10.61 About 200/s.

10.66 100–8,000 Hz, over 40 dB.

10.67 40–14,000 Hz, over 70 dB.

10.72 The former (60 dB SPL) is a bit louder than the latter (59 dB SPL).

Problems of Chapter 11

11.7 (a) (partial answer) 2.5% for the first surface, (b) no.

11.14 7.51 mm, −7.51 mm.

11.15 6.04 mm.

11.17 The retina.
11.22 More damage is done if you look in the direction of the beam because it will focus on the fovea. Damage to the fovea can hurt sharp vision, leaving you with fuzzy vision.

11.23 8.8 cm.

11.34 (partial answer) Refractive index is larger at 630 nm by 0.0008 (if everything else is the same).

11.42 58.62 D (smaller than before), 22.8 mm (longer), mostly due to the smaller refractive index of the crystalline lens.

11.48 $-1.71 \, \text{D}$, $-1.67 \, \text{D}$.

11.53 (c) Myopia. A correction of $-2 \, \text{D}$ would lead to good vision. (The patient has 4 D of accommodation, which is sufficient with this correction.)

11.59 (b) 7.55 mm = 0.00755 m.

11.68 250 lux assuming 500 lumens/W (Fig. 11.49). It is consistent with the levels given in Table 11.5.

Problems of Chapter 12

12.2 110,000 ohms.

12.4 (a) 240 mA, (b) shock and possible ventricular fibrillation would result.

12.11 The large net diffusion of $\text{K}^+$ outside, the impermeability of the membrane to the proteins, which are negatively charged, and the $\text{Na}^+$ pump contribute to the cell being negative relative to the extracellular fluid for a resting axon. The small net diffusion of $\text{Na}^+$ into the cell adds slightly to the positivity inside the cell.

12.19 (partial answer) $6.20 \times 10^{-4} \, \text{s}$ for unmyelinated axons.

12.28 $\sim 80/\text{min}$.

12.30 The dipole usually rotates also. A(c); B(a); C(d); D(b).

Problems of Chapter 13

13.2 Every 17 min.

13.3 (partial answer) 1.6/s for a 1 s delay.
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Appendix D

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