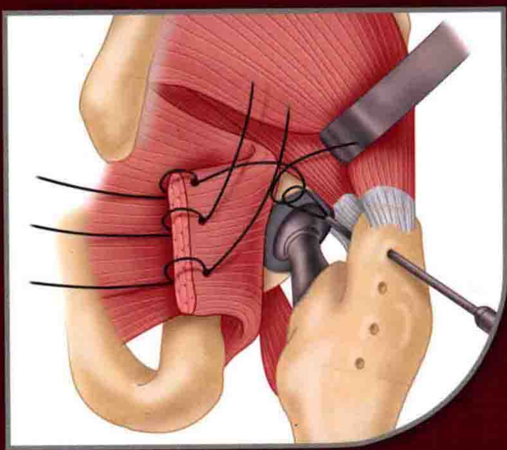
CAMPBELL'S
OPERATIVE ORTHOPAEDICS
TWELFTH EDITION

英文影印版

坎贝尔骨科手术学

第 12 版

成人髋关节置换术分册



S. Terry Canale · James H. Beaty

ELSEVIER

天津出版传媒集团
天津科技翻译出版有限公司

CAMPBELL'S OPERATIVE ORTHOPAEDICS

TWELFTH EDITION

坎贝尔骨科手术学

成人髋关节置换术分册

第 12 版

S. Terry Canale
James H. Beaty

天津出版传媒集团
天津科技翻译出版有限公司

著作权合同登记号：图字：02-2013-138

图书在版编目（CIP）数据

坎贝尔骨科手术学：第12版. 成人髋关节置换术分册：英文/（美）卡奈尔（Canale, S.T.），（美）贝蒂（Beaty, J.H.）主编. —影印本. —天津：天津科技翻译出版有限公司，2013.6

ISBN 978-7-5433-3247-8

Ⅰ.①坎… Ⅱ.①卡… ②贝… Ⅲ.①骨科学—外科手术—英文 ②髋关节置换术—英文 Ⅳ.①R68 ②687.4

中国版本图书馆CIP数据核字（2013）第130640号

This edition of pages 157 through 374 of Campbell's Operative Orthopaedics (12/E) by S. Terry Canale and James H. Beaty is published by arrangement with Elsevier.

ISBN-13:978-0-323-07243-4

ISBN-10:0-323-07243-7

Elsevier(Singapore) Pte Ltd.

3 Killiney Road, #08-01 Winsland House I, Singapore 239519

Tel:(65)6349-0200 Fax:(65)6733-1817

First Published 2013, 2013年初版

授权单位：Elsevier (Singapore) Pte Ltd.
出 版 人：刘 庆
出　　版：天津科技翻译出版有限公司
地　　址：天津市南开区白堤路244号
邮政编码：300192
电　　话：（022）87894896
传　　真：（022）87895650
网　　址：www.tsttpc.com
印　　刷：山东鸿杰印务集团有限公司
发　　行：全国新华书店
版本记录：889×1194　16开本　14.5印张　280千字
2013年6月第1版　2013年6月第1次印刷
定价：78.00元

（如发现印装问题，可与出版社调换）

影印版序

《坎贝尔骨科手术学》由世界级专家联袂编撰，自1939年问世以来，这部巨著伴随了一代又一代骨科医生的成长，成为全球骨科医生不可或缺的参考书，是骨科学领域最权威的著作，同样也被我国广大骨科医生奉为经典。

2013年初，Elsevier 出版公司出版了这部骨科学“圣经”的最新版本——第12版，作为一名旧版的老读者，再次切身感受到该书的严谨、科学。新版分4卷，19部分，89章。介绍了骨科手术的基本原理，详细讲述了髋、膝、踝、肩肘关节置换术，以及截肢与感染、骨肿瘤、先天性异常和发育异常、脊柱损伤、运动损伤、成人骨折与脱位、周围神经损伤、手和足踝部损伤的各种手术技术、儿童神经系统疾病及骨折与脱位。此外，还介绍了关节镜及显微外科的先进手术技术和经验。本书的特点是详细地叙述了各种手术的细节，包括手术指征、手术前后处理和并发症防治的原则、各种技巧和注意事项，还配备详细的手术图解，编排合理，非常符合临床骨科医生的学习需要。

新版《坎贝尔骨科手术学》达到了“去粗存精”、“去伪存真”之目的，删除了第11版中一些陈旧的观点和方法，吸取了近年来的最新成果，除保留作为“金标准”的经典技术之外，还介绍了大量新技术、新装备，并强调了微创骨科技术，对当前及今后一段时间的骨科临床和科研具有非常重要的指导作用。新版配图7000余幅，其中很多图片为重新绘制，直观展现骨科手术技术要点。

随着我国骨科界对外交流的日益增加，以及骨科医生英语水平的整体提高，越来越多的骨科医生希望能够尽快读到原汁原味的国外经典之作，恰逢此时，天津科技翻译出版有限公司在第12版《坎贝尔骨科手术学》刚刚推出之际，便立即引进了这部巨著的影印版本，几乎与原版同步出版，让国内读者在第一时间即能零距离地领略到这部经典原著的风采，更直接地分享这些国际骨科权威专家们对骨科手术学的真知灼见！考虑到读者的需求，出版社将影印版设计为两种形式出版。一种是如原版书，做成精装四卷的形式，另一种则按照骨科学的分支，将这套专著做成平装版，分为14个分册，可以让读者各取所需。此外，影印版均采用优质铜版纸印刷，保持了原版书的风貌，其性价比之高在近些年的影印版书中亦不多见。

最后，借此书出版之际，愿全体骨科同仁不断更新知识、锻炼技能，更好地为广大患者解除病痛，为我国的骨科事业的快速、健康发展做出更大的贡献！

中国工程院院士

PREFACE

As with every edition of this text, we have been amazed by the multitude of new techniques, new equipment, and new information generated by our orthopaedic colleagues worldwide. The emphasis on less-invasive surgical techniques for everything from hallux valgus correction to spine surgery to total joint arthroplasty has produced a variety of new approaches and new devices. The use of arthroscopy and endoscopy continues to expand its boundaries. We have attempted to include the latest orthopaedic procedures, while retaining many of the classic techniques that remain the "gold standards."

Some of the changes in this edition that we believe will make it easier to use include the complete redrawing of the thousands of illustrations, the combining of some chapters and rearrangement of others to achieve a more logical flow of information, the addition of several new chapters, and the placement of references published before 2000 on the website only. Full access to the text and to an increased number of surgical videos is available on Expert-Consult.com, which is included with the purchase of the text. This combination of traditional and electronic formats, we believe, will make this edition of *Campbell's Operative Orthopaedics* easily accessible and useable in any situation, making it easier for orthopaedists to ensure the highest quality of patient care.

The true "heroes" of this work are our dedicated authors, who are willing to endure time away from their families and their practices to make sure that their contributions are as up-to-date and informational as possible. The revision process is lengthy and arduous, and we are truly appreciative of the time and effort expended by all of our contributors. As always, the personnel of the Campbell Foundation—Kay Daugherty, Barry Burns, Linda Jones, and Joan Crowson—were essential in getting the ideas and information from 40 authors into a workable form. The progress of the book was marked by the proliferation of paper-stuffed file folders spread across their offices. Managing to transform all of that raw material into readable text and illustrative images is always an amazing accomplishment. Our thanks, too, to the individuals at Elsevier publishing who provided much guidance, encouragement, and assistance: Taylor Ball, Content Development Editor; Dolores Meloni, Executive Content Strategist; Mary Gatsch, Publishing Director; and John Casey, Project Manager.

We are most grateful to our families, especially our wives, Sissie Canale and Terry Beaty, who patiently endured our total immersion in the publication process.

The individuals who often are overlooked, or at least not recognized often enough, are the community of orthopaedic surgeons to whom we are indebted for their expertise and innovation that make a textbook such as ours necessary. As Dr. Campbell noted in the preface to the first edition of this text, "In some of the chapters we have drawn heavily from authoritative articles on special subjects; the author gratefully acknowledges his indebtedness for this material." We are indeed grateful, and honored and humbled, to be the conduit of such remarkable skill and knowledge that help us to make the most current information available to our readers. We hope that this latest edition of *Campbell's Operative Orthopaedics* will prove to be a valuable tool in providing the best of care to orthopaedic patients.

S. Terry Canale, MD
James H. Beaty, MD

CONTRIBUTORS

WILLIAM E. ALBERS, MD
Assistant Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

FREDERICK M. AZAR, MD
Professor
Director, Sports Medicine Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Chief-of-Staff, Campbell Clinic
Memphis, Tennessee

JAMES H. BEATY, MD
Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

JAMES H. CALANDRUCCIO, MD
Associate Professor
Director, Hand Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

FRANCIS X. CAMILLO, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

S. TERRY CANALE, MD
Harold H. Boyd Professor and Chair
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

DAVID L. CANNON, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

KEVIN B. CLEVELAND, MD
Instructor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

ANDREW H. CRENSHAW, JR., MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

JOHN R. CROCKARELL, JR., MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

GREGORY D. DABOV, MD
Assistant Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

RAYMOND J. GARDOCKI, MD
Instructor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

JAMES L. GUYTON, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

JAMES W. HARKESS, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

ROBERT K. HECK, JR., MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

SUSAN N. ISHIKAWA, MD
Assistant Professor
Co-Director, Foot and Ankle Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

MARK T. JOBE, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

DEREK M. KELLY, MD
Assistant Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

DAVID G. LAVELLE, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

SANTOS F. MARTINEZ, MD
Instructor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

ANTHONY A. MASCIOLI, MD
Assistant Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

MARC J. MIHALKO, MD
Instructor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

WILLIAM W. MIHALKO, MD
Professor, H.R. Hyde Chair of Excellence in Rehabilitation Engineering
Director, Biomedical Engineering
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

ROBERT H. MILLER III, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

G. ANDREW MURPHY, MD
Assistant Professor
Co-Director, Foot and Ankle Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

ASHLEY L. PARK, MD
Clinical Assistant Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

EDWARD A. PEREZ, MD
Associate Professor
Director, Trauma Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

BARRY B. PHILLIPS, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

DAVID R. RICHARDSON, MD
Assistant Professor
Residency Program Director
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

E. GREER RICHARDSON, MD
Professor Emeritus
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

MATTHEW I. RUDLOFF, MD
Assistant Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

JEFFREY R. SAWYER, MD
Associate Professor
Director, Pediatric Orthopaedic Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

THOMAS W. THROCKMORTON, MD
Associate Professor
Assistant Director, Residency Program
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

PATRICK C. TOY, MD
Instructor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

WILLIAM C. WARNER, JR., MD
Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

JOHN C. WEINLEIN, MD
Instructor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

A. PAIGE WHITTLE, MD
Associate Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

KEITH D. WILLIAMS, MD
Associate Professor
Director, Spine Fellowship
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

DEXTER H. WITTE, MD
Clinical Assistant Professor of Radiology
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

GEORGE W. WOOD II, MD
Professor
University of Tennessee–Campbell Clinic
Department of Orthopaedic Surgery and Biomedical Engineering
Memphis, Tennessee

CONTENTS

PART II

RECONSTRUCTIVE PROCEDURES OF THE HIP IN ADULTS

3 Arthroplasty of the Hip 158
James W. Harkess and John R. Crockarell, Jr.

4 Hip Resurfacing 311
David G. Lavelle

5 Arthrodesis of the Hip 325
William E. Albers

6 Hip Pain in the Young Adult and Hip Preservation Surgery 333
James L. Guyton

List of Techniques

Arthroplasty of the Hip

3-1 Preoperative Templating for Total Hip Arthroplasty (Capello), 182
3-2 Total Hip Arthroplasty Through the Posterolateral Approach with Posterior Dislocation of the Hip, 185
3-3 Implantation of Cementless Acetabular Component, 189
3-4 Implantation of Cemented Acetabular Component, 191
3-5 Implantation of Cementless Femoral Component, 193
3-6 Implantation of Cemented Femoral Component, 197
3-7 Transtrochanteric Approach for Revision Total Hip Arthroplasty, 266
3-8 Removal of Cemented Femoral Component, 267
3-9 Removal of a Cementless Femoral Component, 268
3-10 Removal of Implants with Extensive Distal Bone Ingrowth (Glassman and Engh), 269
3-11 Extended Trochanteric Osteotomy (Younger et al.), 270
3-12 Removal of a Broken Femoral Component (Collis and Dubrul), 271
3-13 Removal of a Broken Stem (Moreland, Marder, and Anspach), 272
3-14 Removal of a Broken Stem, 273
3-15 Removal of Femoral Cement, 273
3-16 Removal of Distal Cement with a High-Speed Burr (Turner et al.), 275
3-17 Removal of Distal Cement with a High-Speed Burr (Mallory), 276
3-18 Removal of a Loose All-Polyethylene Cup, 277
3-19 Removal of a Metal-Backed Cemented Acetabular Component, 277
3-20 Removal of a Cementless Acetabular Component (Mitchell), 279
3-21 Management of Acetabular Cavitary Deficits, 282
3-22 Management of Segmental Acetabular Deficit, 284
3-23 Management of Segmental Acetabular Deficit (Fletcher), 285
3-24 Management of Combined Deficits with Structural Grafting (Sporer et al.), 286
3-25 Management of Pelvic Discontinuity with Antiprotrusio Cage (Winter et al.), 287
3-26 Management of Pelvic Discontinuity with Allografting and Custom Component (DeBoer et al.), 288
3-27 Management of Femoral Deficit with Modular Femoral Component (Cameron), 293
3-28 Revision with Extensively Porous-Coated Femoral Stem (Mallory and Head), 294
3-29 Management of Proximal Femoral Bone Loss with Distally Fixed Stem (Kwong et al.), 294
3-30 Management of Proximal Femoral Deficiencies with Impaction Bone Grafting and Cemented Revision Stem (Gie, Modified), 295
3-31 Management of Massive Deficits with Proximal Femoral Allograft-Prosthesis Composite, 297
3-32 Management of Massive Deficits with Modular Megaprosthesis (Klein et al.), 298

Arthrodesis of the Hip

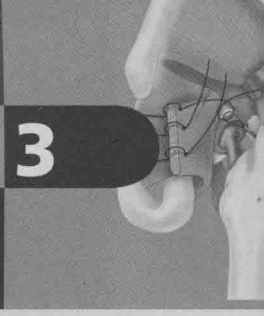
5-1 Arthrodesis with Cancellous Screw Fixation (Benaroch et al.), 326
5-2 Arthrodesis with Cobra Plate Fixation (Murrell and Fitch), 326
5-3 Arthrodesis with Hip Compression Screw Fixation (Pagnano and Cabanela), 328
5-4 Arthrodesis in the Absence of the Femoral Head (Abbott, Fischer, and Lucas), 328
5-5 Arthrodesis of the Proximal Femur to the Ischium (Bosworth), 329
5-6 Arthrodesis with Cobra-Head Plate Fixation After Failed Total Hip Arthroplasty (Kostuik and Alexander), 330

Hip Pain in the Young Adult and Hip Preservation Surgery

6-1 Surgical Dislocation of the Hip (Ganz et al.), 343
6-2 Combined Hip Arthroscopy and Limited Open Osteochondroplasty (Clohisy and McClure), 345
6-3 Bernese Periacetabular Osteotomy (Matheney et al.), 352
6-4 Osteonecrosis of the Femoral Head: Core Decompression (Hungerford), 361
6-5 Osteonecrosis of the Femoral Head: Core Decompression—Percutaneous Technique (Mont et al.), 361
6-6 Osteonecrosis of the Femoral Head: Vascularized Fibular Grafting (Urbaniak et al.), 363

ARTHROPLASTY OF THE HIP

CHAPTER 3



James W. Harkess • John R. Crockarell, Jr.

APPLIED BIOMECHANICS 159
FORCES ACTING ON THE HIP 159
STRESS TRANSFER TO BONE 161
DESIGN AND SELECTION OF TOTAL HIP COMPONENTS 163
FEMORAL COMPONENTS 164
Cemented Femoral Components 166
Cementless Femoral Components 168
Specialized and Custom-Made Femoral Components 172
ACETABULAR COMPONENTS 172
Cemented Acetabular Components 172
Cementless Acetabular Components 173
Alternative Bearings 175
INDICATIONS FOR TOTAL HIP ARTHROPLASTY 178
CONTRAINDICATIONS TO TOTAL HIP ARTHROPLASTY 179
PREOPERATIVE EVALUATION 180
PREOPERATIVE RADIOGRAPHS 181
PREPARATION AND DRAPING 183
SURGICAL APPROACHES AND TECHNIQUES 184
IMPLANTATION OF CEMENTLESS ACETABULAR COMPONENTS 188
IMPLANTATION OF CEMENTED ACETABULAR COMPONENTS 191
IMPLANTATION OF CEMENTLESS FEMORAL COMPONENTS 193
IMPLANTATION OF CEMENTED FEMORAL COMPONENTS 197
MINIMALLY INVASIVE TECHNIQUES 200
COMPUTER-ASSISTED SURGERY 203
TROCHANTERIC OSTEOTOMY 203
SURGICAL PROBLEMS RELATIVE TO SPECIFIC HIP DISORDERS 207
ARTHRITIC DISORDERS 207
Osteoarthritis (Primary or Secondary Hypertrophic Arthritis or Degenerative Arthritis) 207
Rheumatoid Arthritis 208
OSTEONECROSIS 208
PROTRUSIO ACETABULI 209
DEVELOPMENTAL DYSPLASIA 211
DWARFISM 217
POSTTRAUMATIC DISORDERS 217
Acute Femoral Neck Fractures 217
Failed Hip Fracture Surgery 220
Acetabular Fractures 221
FAILED RECONSTRUCTIVE PROCEDURES 222
Proximal Femoral Osteotomy and Deformity 222
Acetabular Osteotomy 223
Arthrodesis and Ankylosis 224
METABOLIC DISORDERS 225
Paget Disease 225
Gaucher Disease 226
Sickle Cell Anemia 226
Chronic Renal Failure 227
Hemophilia 227
INFECTIOUS DISORDERS 227
Pyogenic Arthritis 227
Tuberculosis 228
TUMORS 228
NEUROMUSCULAR DISORDERS 228
COMPLICATIONS 228
MORTALITY 228
HEMATOMA FORMATION 229
HETEROTOPIC OSSIFICATION 229
THROMBOEMBOLISM 230
NERVE INJURIES 233
VASCULAR INJURIES 234
LIMB-LENGTH DISCREPANCY 235
DISLOCATION 237
FRACTURES 241
TROCHANTERIC NONUNION 248
INFECTION 250
Antibiotic Prophylaxis 250
Classification 251
Diagnosis 251
Management 252
Reconstruction After Infection 254
LOOSENING 255
Femoral Loosening 256
Acetabular Loosening 259
Diagnosis 260
OSTEOLYSIS 261
REVISION OF TOTAL HIP ARTHROPLASTY 264
INDICATIONS AND CONTRAINDICATIONS 264
PREOPERATIVE PLANNING 265
SURGICAL APPROACH 265
Removal of Femoral Cement 273
Removal of a Cemented Acetabular Component 277
RECONSTRUCTION OF ACETABULAR DEFICIENCIES 280
Classification 280
Management 282
Segmental Deficits 283
Combined Deficits 286
Pelvic Discontinuity 286
RECONSTRUCTION OF FEMORAL DEFICIENCIES 288
Classification 288
Management 290
Segmental Deficits 290
Cavitary Deficits 292
Massive Deficits 296
Femoral Deformity 299
POSTOPERATIVE MANAGEMENT OF TOTAL HIP ARTHROPLASTY 300

Total hip arthroplasty is the most commonly performed adult reconstructive hip procedure. This chapter discusses cemented and noncemented arthroplasties, bearing choices, and current trends in minimally invasive techniques. In addition, revision hip arthroplasty, which comprises an enlarging segment of procedures performed, is reviewed.

The results of the Charnley total hip arthroplasty are the benchmark for evaluating the performance of other arthroplasties. The laboratory and clinical contributions of Charnley have improved the quality of life for many patients. Nevertheless, the history of hip arthroplasty has been dynamic, and research continues to improve results, especially in young patients. Investigation has proceeded along multiple paths, including (1) improvement in the durability of implant fixation, (2) reduction in the wear of the articulating surfaces, and (3) technical modifications in the operation to speed rehabilitation and reduce implant-positioning errors.

In response to the problem of loosening of the stem and cup based on the alleged failure of cement, press-fit, porous-coated, and hydroxyapatite-coated stems and cups have been investigated as ways to eliminate the use of cement and to use bone ingrowth or ongrowth as a means of achieving durable skeletal fixation. Although some initial cementless implant designs have proved very successful, others have been beset by premature and progressive failure because of inadequate initial fixation, excessive wear, and periprosthetic bone loss secondary to particle-induced osteolysis. As experience has accumulated, the importance of certain design parameters has become apparent and the use of cementless fixation for the femoral and acetabular components has become more common.

Many different techniques have evolved to improve cemented femoral fixation, including injection of low-viscosity cement, occlusion of the medullary canal, reduction of porosity, pressurization of the cement, and centralization of the stem. Similar techniques have been less successful in improving the results of acetabular fixation. Stem fracture has been largely eliminated by routine use of superalloys in their fabrication.

As technological advances improve the longevity of implant fixation, problems related to wear of articulating surfaces have emerged. Highly crosslinked polyethylenes have demonstrated reduced wear and have now largely replaced conventional ultra-high-molecular-weight polyethylene. Ceramic-ceramic and metal-metal articulations have been used because of their low coefficient of friction and superior in vitro wear characteristics, although with less favorable results. Additionally, the use of these more wear-resistant bearings has led to the use of larger component head sizes and modifications of postoperative regimens.

It is important to consider the problems of previous materials and design modifications that did not become apparent until the results of a sufficient number of 5-year or more follow-up studies were available. There is little debate that the results of revision procedures are less satisfactory and that primary total hip arthroplasty offers the best chance of success. Selection of the appropriate patient, the proper implants, and the technical performance of the operation are of paramount importance.

Total hip arthroplasty procedures require the surgeon to be familiar with the many technical details of the operation. To contend successfully with the many problems that occur and to evaluate new concepts and implants, a working knowledge of biomechanical principles, materials, and design also is necessary.

APPLIED BIOMECHANICS

The biomechanics of total hip arthroplasty are different from those of the screws, plates, and nails used in bone fixation because these latter implants provide only partial support and only until the bone unites. Total hip components must withstand many years of cyclic loading equal to at least three times body weight. A basic knowledge of the biomechanics of the hip and of total hip arthroplasty is necessary to perform the procedure properly, to manage the problems that may arise during and after surgery successfully, to select the components intelligently, and to counsel patients concerning their physical activities.

FORCES ACTING ON THE HIP

To describe the forces acting on the hip joint, the body weight can be depicted as a load applied to a lever arm extending from the body's center of gravity to the center of the femoral head (Fig. 3-1). The abductor musculature, acting on a lever arm extending from the lateral aspect of the greater trochanter to the center of the femoral head, must exert an equal moment to hold the pelvis level when in a one-legged stance and a greater moment to tilt the pelvis to the same side when walking. Because the ratio of the length of the lever arm of the body weight to that of the abductor musculature is about 2.5:1, the force of the abductor muscles must approximate 2.5 times the body weight to maintain the pelvis level when standing on one leg. The estimated load on the femoral head in the stance phase of gait is equal to the sum of the forces created by the abductors and the body weight and has been calculated to be three times the body weight; the load on the femoral head during straight-leg raising is estimated to be about the same.

An integral part of the Charnley concept of total hip arthroplasty was to shorten the lever arm of the body weight by deepening the acetabulum and to lengthen the lever arm of the abductor mechanism by reattaching the osteotomized greater trochanter laterally. The moment produced by the body weight is decreased, and the counterbalancing force that the abductor mechanism must exert is decreased. The abductor lever arm may be shortened in arthritis and other hip disorders in which part or all of the head is lost or the neck is shortened. It also is shortened when the trochanter is located posteriorly, as in external rotational deformities, and in many patients with developmental dysplasia of the hip. In an arthritic hip, the ratio of the lever arm of the body weight to that of the abductors may be 4:1. The lengths of the two lever arms can be surgically changed to make their ratio approach 1:1 (see Fig. 3-1). Theoretically, this reduces the total load on the hip by 30%.

It is important to understand the benefits derived from medializing the acetabulum and lengthening the abductor lever arm; however, neither technique is currently emphasized. The principle of medialization has given way to preserving subchondral bone in the pelvis and to deepening the

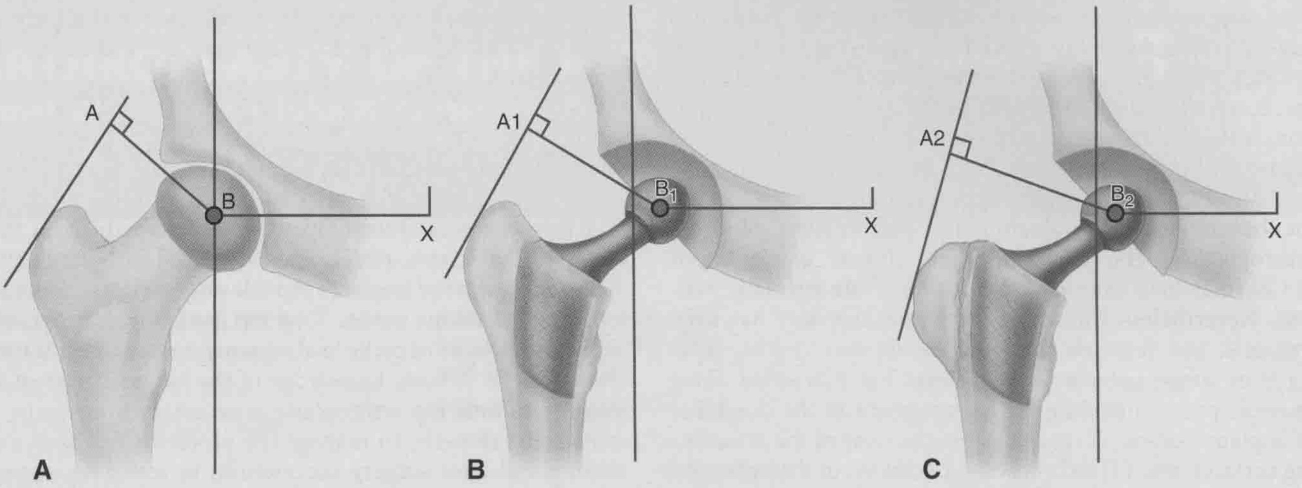


FIGURE 3-1 Lever arms acting on hip joint. **A,** Moment produced by body weight applied at body's center of gravity, X, acting on lever arm, B-X, must be counterbalanced by moment produced by abductors, A, acting on shorter lever arm, A-B. Lever arm A-B may be shorter than normal in arthritic hip. **B,** Medialization of acetabulum shortens lever arm B-X, and use of high offset neck lengthens lever arm A-B. **C,** Lateral and distal reattachment of osteotomized greater trochanter lengthens lever arm A-B further and tightens abductor musculature.

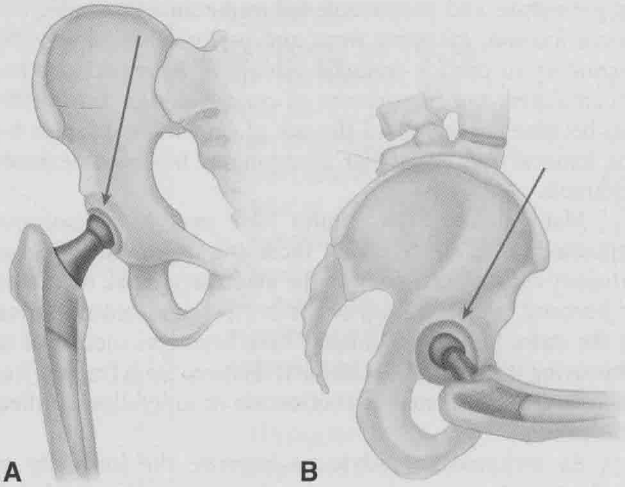


FIGURE 3-2 Forces producing torsion of stem. Forces acting on hip in coronal plane **(A)** tend to deflect stem medially, and forces acting in sagittal plane **(B)**, especially with hip flexed or when lifting, tend to deflect stem posteriorly. Combined, they produce torsion of stem.

acetabulum only as much as necessary to obtain bony coverage for the cup. Because most total hip procedures are now done without osteotomy of the greater trochanter, the abductor lever arm is altered only relative to the offset of the head to the stem. These compromises in the original biomechanical principles of total hip arthroplasty have evolved to obtain beneficial tradeoffs of a biological nature—to preserve pelvic bone, especially subchondral bone, and to avoid problems related to reattachment of the greater trochanter.

Calculated peak contact forces across the hip joint during gait range from 3.5 to 5.0 times the body weight and up to 6 times the body weight during single-limb stance. Experimentally measured forces around the hip joint using instrumented prostheses generally are lower than the forces predicted by analytical models, in the range of 2.6 to 3.0 times the body weight during single-limb stance phase of gait. When lifting, running, or jumping, however, the load may be equivalent to 10 times the body weight. Excess body weight and increased physical activity add significantly to the forces that act to loosen, bend, or break the stem of a femoral component.

The forces on the joint act not only in the coronal plane but, because the body's center of gravity (in the midline anterior to the second sacral vertebral body) is posterior to the axis of the joint, also in the sagittal plane to bend the stem posteriorly. The forces acting in this direction are increased when the loaded hip is flexed, as when arising from a chair, ascending and descending stairs or an incline, or lifting (Fig. 3-2). During the gait cycle, forces are directed against the prosthetic femoral head from a polar angle between 15 and 25 degrees anterior to the sagittal plane of the prosthesis. During stair climbing and straight-leg raising, the resultant force is applied at a point even farther anterior on the head. Such forces cause posterior deflection or retroversion of the femoral component. These so-called out-of-plane forces have been measured at 0.6 to 0.9 times body weight.

Implanted femoral components must withstand substantial torsional forces even in the early postoperative period. Consequently, femoral components used without cement must be designed and implanted so that they are immediately rotationally stable within the femur. Similarly, the shape of a cemented implant must impart rotational stability within its cement mantle.

The location of the center of rotation of the hip from superior to inferior also affects the forces generated around the implant. In a mathematical model, the joint reaction force was lower when the hip center was placed in the anatomical location compared with a superior and lateral or posterior position. Isolated superior displacement without lateralization produces relatively small increases in stresses in the periacetabular bone. This has clinical importance in the treatment

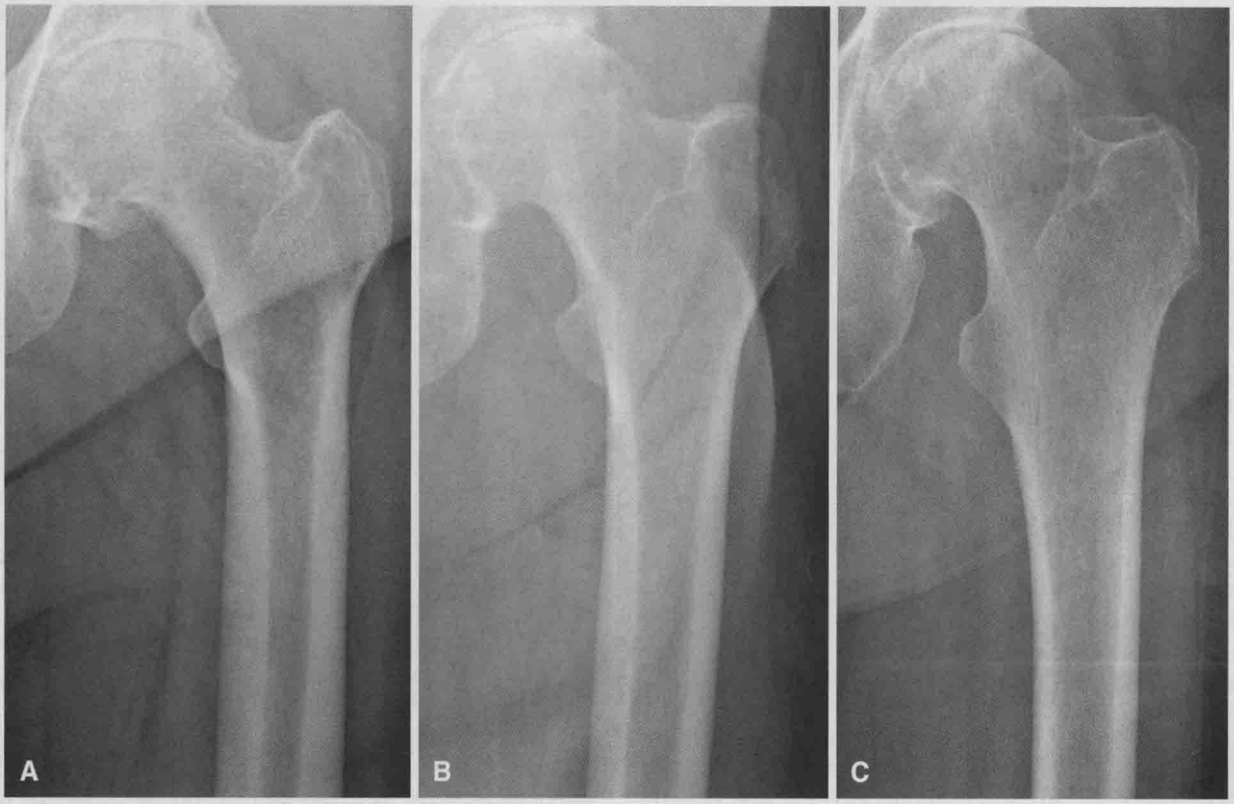


FIGURE 3-3 Radiographic categorization of proximal femurs according to shape; correlation with cortical thickness and canal dimension. (From Dorr LD, Faugere MC, Mackel AM, et al: Structural and cellular assessment of bone quality of proximal femur, *Bone* 14:231, 1993.)

of developmental dysplasia and in revision surgery when superior bone stock is deficient. Placement of the acetabular component in a slightly cephalad position allows improved coverage or contact with viable bone. Nonetheless, clinical studies have documented a higher incidence of progressive radiolucencies and migration of components in patients with protrusion, dysplasia, and revision situations when the hip center was placed in a nonanatomical position.

STRESS TRANSFER TO BONE

The quality of the bone before surgery is a determinant in the selection of the most appropriate implant, optimal method of fixation, response of the bone to the implant, and ultimate success of the arthroplasty. Dorr et al. proposed a radiographic categorization of proximal femurs based on their shape and correlated those shapes with measurements of cortical thickness and canal dimensions (Fig. 3-3). Type A femurs have thick cortices on the anteroposterior view and a large posterior cortex seen on the lateral view. The narrow distal canal gives the proximal femur a pronounced funnel shape or "champagne flute" appearance. The type A femur is more commonly found in men and younger patients and permits good fixation of either cemented or cementless stems. Type B femurs exhibit bone loss from the medial and posterior cortices, resulting in increased width of the intramedullary canal. The shape of the femur is not compromised, and implant fixation is not a problem. Type C femurs have lost much of the medial and posterior cortex. The intramedullary canal diameter is very wide, particularly on the lateral radiograph. The "stovepipe"-shaped type C bone is typically found in older postmenopausal women and creates a less favorable environment for implant fixation.

The material a stem is made of, the geometry and size of the stem, and the method and extent of fixation dramatically alter the pattern in which stress is transferred to the femur. Adaptive bone remodeling arising from stress shielding compromises implant support and predisposes to fracture of the femur or the implant itself. Stress transfer to the femur is desirable because it provides a physiological stimulus for maintaining bone mass and preventing disuse osteoporosis. A decrease in the modulus of elasticity of a stem decreases the stress in the stem and increases stresses to the surrounding bone. This is true of stems made of metals with a lower modulus of elasticity, such as a titanium alloy, if the cross-sectional diameter is relatively small. Larger-diameter stems made of the same material are stronger, but they also are stiffer or less elastic, and the increased cross-sectional diameter negates any real benefits of the lower modulus of elasticity. The bending stiffness of a stem is proportional to the fourth power of the diameter, and small increases in stem diameter produce much larger increments of change in flexural rigidity. When the stem has been fixed within the femur by bone ingrowth, load is preferentially borne by the stiffer structure and the bone of the proximal femur is relieved of stress.

Detailed examinations of stress shielding of the femur after cementless total hip replacement found that almost all femurs showing moderate or severe proximal resorption

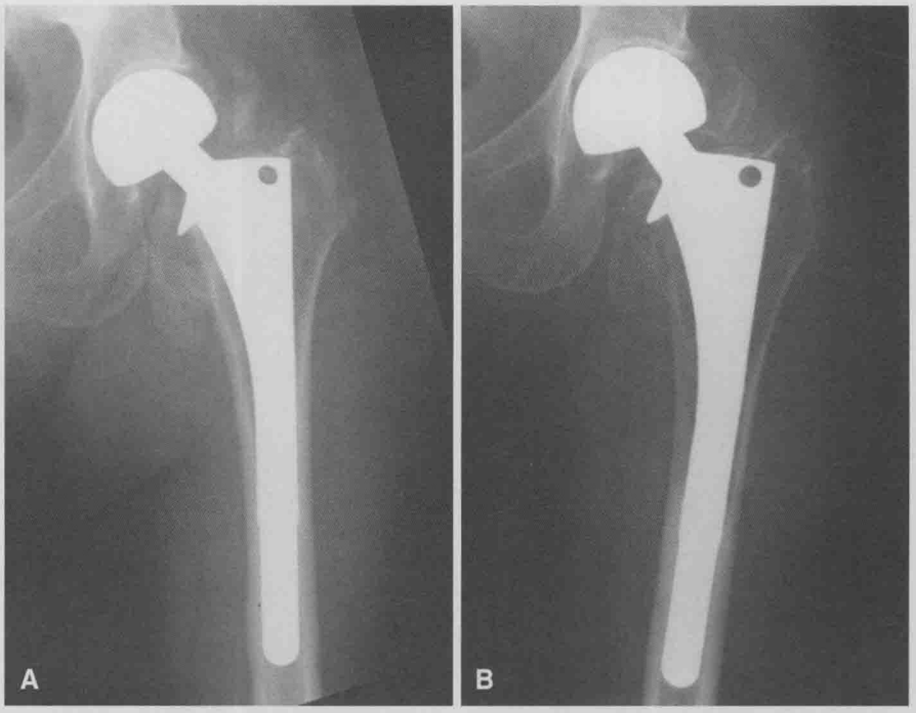


FIGURE 3-4 Response of bone to load. **A,** Postoperative radiograph of extensively porous-coated stem. **B,** Two years later, cortical and cancellous bone density in proximal femur has decreased as a result of stress shielding.

involved stems 13.5 mm in diameter or larger. With a press-fit at the isthmus and radiographic evidence of bone ingrowth, more stress shielding was evident. Extensive porous coating in smaller size stems does not seem to produce severe stress shielding. More recent follow-up with larger stem sizes shows greater stress shielding, however, with more extensively coated stems (Fig. 3-4). Localized bone hypertrophy can be seen in areas where an extensively porous-coated stem contacts the cortex. This is seen often at the distal end of the porous coating with an extensively coated stem. Such hypertrophy is less pronounced when the porous surface is confined to the proximal portion of the stem.

Using videodensitometry to analyze autopsy-retrieved femurs, Maloney et al. found that for cemented and cementless implants, the area of greatest decrease in bone mineral density occurred in the proximal medial cortex. If a prosthesis has a collar that is seated on the cut surface of the neck, it is postulated that axial loading of the bone would occur in this area. It is technically difficult, however, to obtain this direct contact of collar or cement with the cut surface of bone. Although the role of a collar in preventing loosening of a cemented femoral component has not been clearly established, any loading of the proximal medial neck is likely to decrease bone resorption and reduce stresses in the proximal cement. A collar also serves as a simple means of determining the depth of insertion of the femoral component because vision is temporarily obscured by extrusion of the cement. The presence of a collar on cementless femoral components is more controversial because it may prevent complete seating of the stem, making it loose at implantation.

Cementless stems generally produce strains in the bone that are more physiological than the strains caused by fully cemented stems, depending on the stem size and the extent of porous coating. Proximal medial bone strains have been found to be 65% of normal with a collarless press-fit stem and 70% to 90% with a collared stem with an exact proximal fit. A loose-fitted stem with a collar can produce proximal strains greater than in the intact femur, although the consequences of a loose stem negate any potential benefits in loading provided by the collar. When a stem is loaded, it produces circumferential or hoop stresses in the proximal femur. Proximal wedging of a collarless implant may generate excessive hoop strains that cause intraoperative and postoperative fractures of the proximal femur.

Stem shape also seems to affect stress transfer to bone. In a review of three different types of titanium stems with tapered geometries, Mallory, Head, and Lombardi found an overall incidence of radiographic proximal femoral bone atrophy of only 6% of 748 arthroplasties studied. In no patient was the proximal bone loss as severe as that seen in patients with stems of a cylindrical distal geometry that filled the diaphysis.

Cadaver studies have identified a wide variability in the degree and location of bone remodeling between individuals in clinically successful arthroplasties with solid fixation. A strong correlation was shown, however, between the bone mineral density in the opposite femur and the percentage of mineral loss in the femur that had been operated on, regardless of the method of implant fixation; it seems that patients with diminished bone mineral density before surgery are at greatest risk for significant additional bone loss after cemented and cementless total hip arthroplasty.

The amount of stress shielding that is acceptable in the clinical setting is difficult to determine. A point of equilibrium is reached, and bone loss does not often progress after 2 years. In a series of 208 hip arthroplasties followed for a mean 13.9 years, Engh et al. reported patients with radiographically evident stress shielding had lower mean walking scores but no increase in other complications and were less likely to require revision for stem loosening or osteolysis. Although proximal femoral stress shielding does not seem to affect adversely early or midterm clinical results, experience with failed cemented implants also has shown that revision surgery becomes more complex when femoral bone stock has been lost. Ongoing investigations into materials of lower modulus of elasticity and stem geometries that diminish flexural rigidity are likely to be beneficial in reducing adverse femoral remodeling.

On the pelvic side, finite analysis has indicated that with the use of a cemented polyethylene cup, peak stresses develop in the pelvic bone. A metal-backed cup with a polyethylene liner reduces the high areas of stress and distributes the stresses more evenly. Similar studies have indicated that increased peak stresses develop in the trabecular bone when the subchondral bone is removed and that decreased peak stresses develop when a metal-backed component is used. The highest stresses in the cement and trabecular bone develop when a thin-walled, polyethylene acetabular component is used and when the subchondral bone has been removed. A thick-walled polyethylene cup of 5 mm or more, as opposed to a thin-walled polyethylene cup, tends to reduce the stresses in the trabecular bone, similar to the effect of the metal-backed cup. The preservation of subchondral bone in the acetabulum and the use of a metal-backed cup or thick-walled polyethylene cup decrease the peak stress levels in the trabecular bone of the pelvis.

Favorable early results with metal-backed, cemented acetabular components led to their widespread use in the past. Longer follow-up has shown no sustained benefit, however, from the use of metal backing, and in some series survivorship of the cemented metal-backed acetabular components has been worse than that of components without metal backing. Using a thick-walled, all-polyethylene component and retaining the subchondral bone of the acetabulum are two steps that seem to provide a satisfactory compromise without excessive stress shielding or stress concentration.

When cementless acetabular fixation is used, metal backing is required for skeletal fixation. Ideally, the metal should contact acetabular subchondral bone over a wide area to prevent stress concentration and to maximize the surface area available for biological fixation. The accuracy of acetabular preparation and the shape and size of the implant relative to the prepared cavity dramatically affect this initial area of contact and the transfer of stress from implant to the pelvis. If a hemispherical component is slightly undersized relative to the acetabulum, stress is transferred centrally over the pole of the component, with the potential for peripheral gaps between the implant and bone. Conversely, if the component is slightly larger than the prepared cavity, stress transfer occurs peripherally, with the potential for fracture of the acetabular rim during implantation (see section on implantation of cementless acetabular components). Polar gaps also may remain from incomplete seating of the component.

The manner of stress transfer from a cementless acetabular component to the surrounding acetabular bone dictates its initial stability. As the cup is impacted into the acetabulum, forces generated by elastic recoil of the bone stabilize the implant. Peripheral strains acting on a force vector perpendicular to the tangent at the rim stabilize the cup. Strains medial to the rim generate a force vector that pushes laterally and destabilizes the cup (Fig. 3-5).

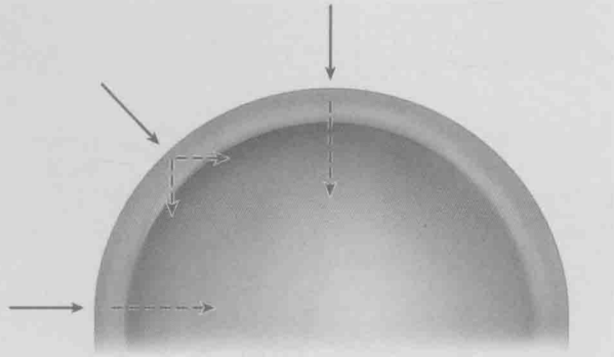


FIGURE 3-5 Destabilization of cup from strains medial to rim.

Stress shielding of the periacetabular bone by cementless implants has received less attention than with femoral components but does occur. Using a novel method of CT-assisted osteodensitometry, Mueller et al. assessed bone density around cementless titanium acetabular components at 10 days and 1 year postoperatively. Cortical bone density cephalad to the implant increased by 3.6%. Conversely, cancellous bone density decreased by 18%, with the area of greatest loss anterior to the cup. The clinical importance of acetabular stress shielding has not been determined.

DESIGN AND SELECTION OF TOTAL HIP COMPONENTS

Total hip femoral and acetabular components of various materials and a multitude of designs are currently available. Few implant designs prove to be clearly superior or inferior to others. Certain design features of a given implant may provide an advantage in selected situations. Properly selected and implanted total hip components of most designs can be expected to yield satisfactory results in a high percentage of patients. No implant design or system is appropriate for every patient, and a general knowledge of the variety of component designs and their strengths and weaknesses is an asset to the surgeon. Selection is based on the patient's needs, the patient's anticipated longevity and level of activity, the bone quality and dimensions, the ready availability of implants and proper instrumentation, and the experience of the surgeon.

We routinely use many total hip systems from different manufacturers; we present here an overview of the available systems, emphasizing similar and unique features. Numerous investigators and manufacturers have changed their designs within a relatively short time to incorporate newer concepts, and this confuses many orthopaedic surgeons and patients. The surgeon's recommendations should be tempered by the knowledge that change does not always bring about improvement and that radical departure from proven concepts of implant design yields unpredictable long-term results.

Total hip femoral and acetabular components are commonly marketed together as a total hip system. Although these systems are often convenient, the variety of modular head sizes with most femoral components allows use with

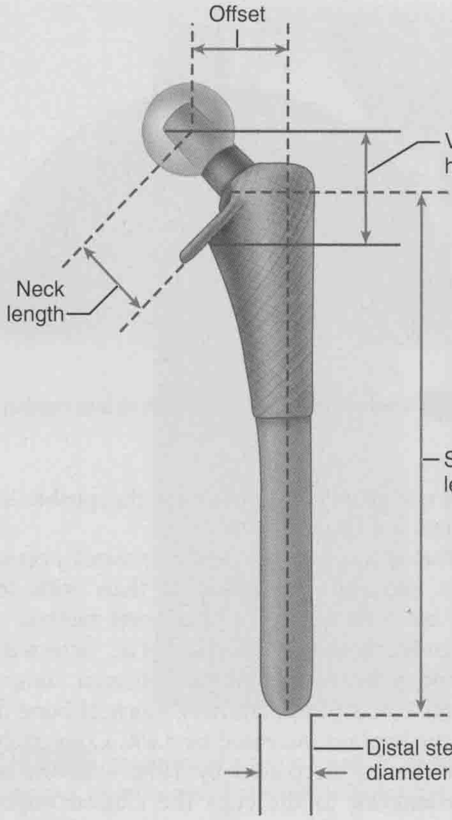


FIGURE 3-6 Features of femoral component. Neck length is measured from center of head to base of collar; head-stem offset, from center of head to line through axis of distal part of stem; stem length, from medial base of collar to tip of stem; and angle of neck, by intersection of line through center of head and neck with another along lateral border of distal half of stem. Platform is medial extension of collar.

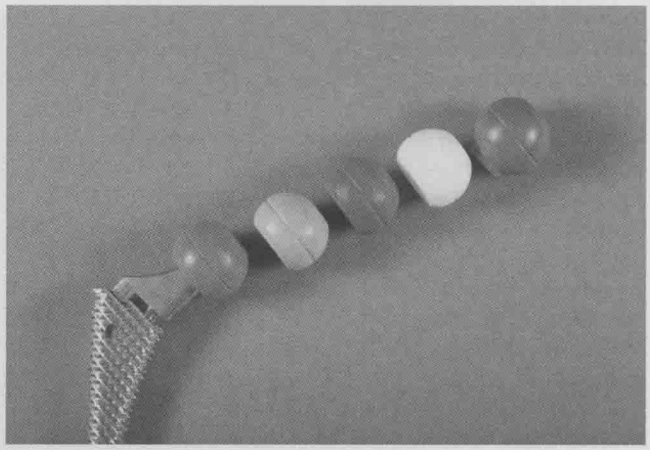


FIGURE 3-7 Modular heads for femoral components. Neck taper mates with modular femoral heads. Motion is absent between head and neck segments. Different diameter heads with various neck extensions are available. Extended neck, or "skirt," of longer components has larger diameter than neck of conventional components, and arc of motion of hip is decreased.

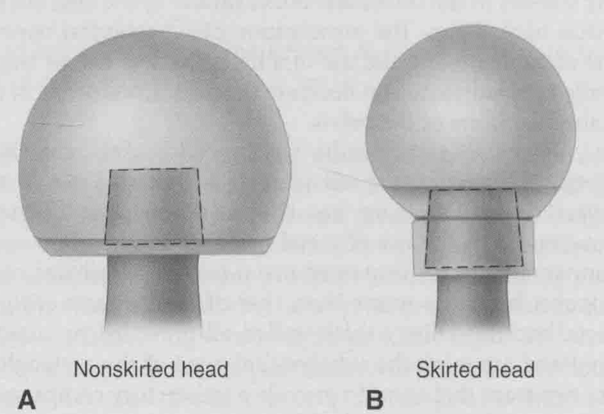


FIGURE 3-8 Head-to-neck ratio of implants. Large-diameter head with trapezoidal neck **(A)** has greater range of motion and less impingement than smaller diameter head and skirted modular neck **(B)**.

other types of acetabular components if necessary. Femoral and acetabular components are discussed separately.

FEMORAL COMPONENTS

The primary function of the femoral component is the replacement of the femoral head and neck after resection of the arthritic or necrotic segment. The ultimate goal of a biomechanically sound, stable hip joint is accomplished by careful attention to restoration of the normal center of rotation of the femoral head. This location is determined by three factors: (1) vertical height (vertical offset), (2) medial offset (horizontal offset or, simply, offset), and (3) version of the femoral neck (anterior offset) (Fig. 3-6). Vertical height and offset increase as the neck is lengthened, and proper reconstruction of both features is the goal when selecting the length of the femoral neck. In most modern systems, neck length is adjusted by using modular heads with variable internal recesses that fit onto a Morse taper on the neck of the stem (Fig. 3-7). Neck length typically ranges from 25 to 50 mm, and adjustment of 8 to 12 mm for a given stem size routinely is available. When a long neck length is required for a head diameter up to 32 mm, a skirt extending from the lower aspect of the head may be required to fully engage the Morse taper (Fig. 3-8). For heads larger than 32 mm a skirt is unnecessary even for longer neck lengths.

Vertical height (vertical offset) is determined primarily by the base length of the prosthetic neck plus the length gained by the modular head used. In addition, the depth the implant is inserted into the femoral canal alters vertical height. When cement is used, the vertical height can be adjusted further by variation in the level of the femoral neck osteotomy. This additional flexibility may be unavailable when a cementless femoral component is used because depth of insertion is determined more by the fit within the femoral metaphysis than by the level of the neck osteotomy.

Offset (i.e., horizontal offset) is the distance from the center of the femoral head to a line through the axis of the distal part of the stem and is primarily a function of stem design. Inadequate restoration of offset shortens the moment arm of the abductor musculature and results in increased

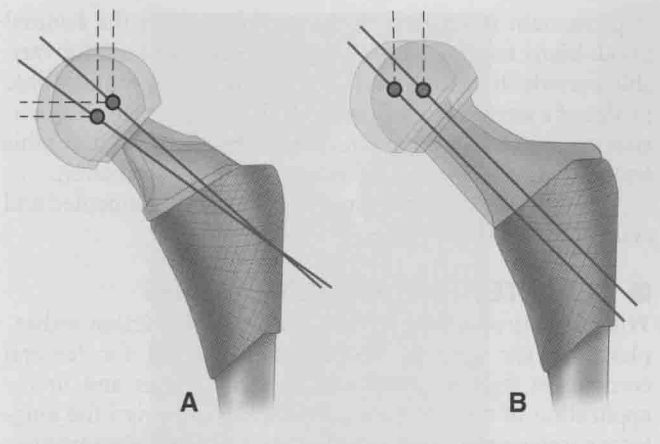


FIGURE 3-9 Variations in femoral component necks to increase offset. **A,** Neck-stem angle is reduced. **B,** Neck is attached at more medial position on stem.

joint reaction force, limp, and bone impingement, which may result in dislocation. Offset can be increased by simply using a longer modular neck, but doing so also increases vertical height, which may result in overlengthening of the limb. To address individual variations in femoral anatomy, many components are now manufactured with standard and high offset versions. This is accomplished by reducing the neck-stem angle (typically to about 127 degrees) or by attaching the neck to the stem in a more medial position (Fig. 3-9). In this manner, offset is increased without limb lengthening.

Version refers to the orientation of the neck in reference to the coronal plane and is denoted as *anteversion* or *retroversion.* Restoration of femoral neck version is important in achieving stability of the prosthetic joint. The normal femur has 10 to 15 degrees of anteversion of the femoral neck in relation to the coronal plane when the foot faces straight forward, and the prosthetic femoral neck should approximate this. Proper neck version usually is accomplished by rotating the component within the femoral canal. This presents little problem when cement is used for fixation; however, when press-fit fixation is used, the femoral component must be inserted in the same orientation as the femoral neck to maximize the fill of the proximal femur and achieve rotational stability of the implant. This problem can be circumvented by the use of a modular femoral component in which the stem is rotated independent of the metaphyseal portion. So-called anatomical stems have a slight proximal posterior bow to reproduce the contour of the femoral endosteum, predetermining the rotational alignment of the implant. Most such stems have a few degrees of anteversion built into the neck to compensate for this, and separate right and left stems are required. Finally, newer femoral components with completely modular necks in different geometries and lengths allow the adjustment of length, offset, and version independently (Fig. 3-10).

The size of the femoral head, the ratio of head and neck diameters, and the shape of the neck of the femoral component have a substantial effect on the range of motion of the hip, the degree of impingement between the neck and rim of the socket, and the stability of the articulation. This impingement can lead to dislocation, accelerated polyethylene wear, acetabular component loosening, and liner dislodgment or fracture. For a given neck diameter, the use of a larger femoral

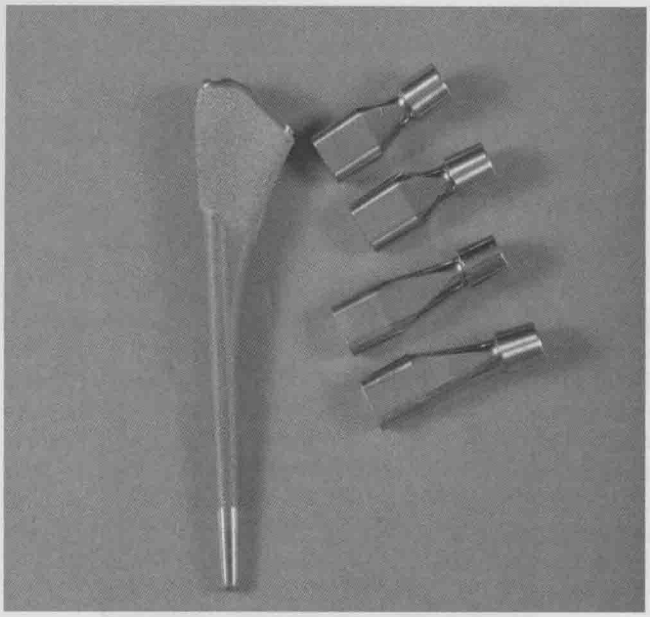


FIGURE 3-10 Modular femoral neck with taper junctions for stem body and femoral head. Multiple configurations allow independent adjustment of length and offset and version.

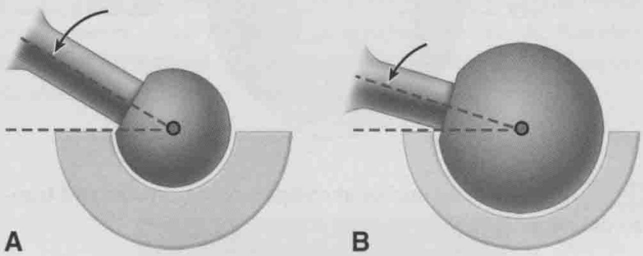


FIGURE 3-11 Range of motion with different head sizes. For given diameter neck, implant with smaller femoral head **(A)** will have lesser arc of motion than larger one **(B)**.

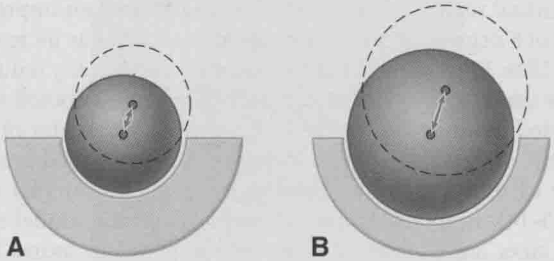


FIGURE 3-12 Jump distance. With subluxation, smaller head **(A)** has shorter distance to travel before escaping rim of acetabular component than larger one **(B)**.

head increases the head-neck ratio and the range of motion before the neck impinges on the rim of the socket will be greater (Fig. 3-11). When this impingement does occur, the femoral head is levered out of the socket. The "jump distance" is the distance the head must travel to escape the rim of the socket and is generally approximated to be half the diameter of the head (Fig. 3-12). For both of these reasons, a larger-diameter head is theoretically more stable than a smaller one. The introduction of advanced bearing surfaces has allowed

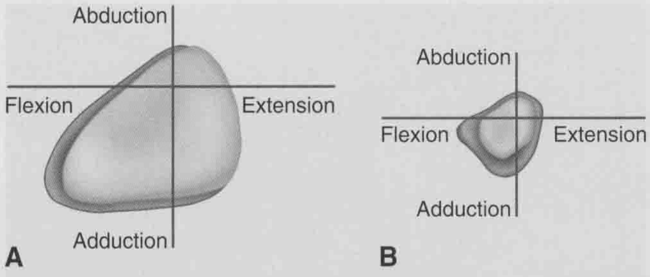


FIGURE 3-13 Effects of head size and neck geometry on range of motion. **A,** Changing from 28-mm head *(dark shading)* to 32-mm head *(light shading)* results in 8-degree increase in flexion before impingement. **B,** Large circular taper has dramatically decreased range of motion to impingement *(light shading),* which is diminished even further by having skirted modular head *(dark shading).* (From Barrack RL, Lavernia C, Ries M, et al: Virtual reality computer animation of the effect of component position and design on stability after total hip arthroplasty, *Orthop Clin North Am* 32:569, 2001.)

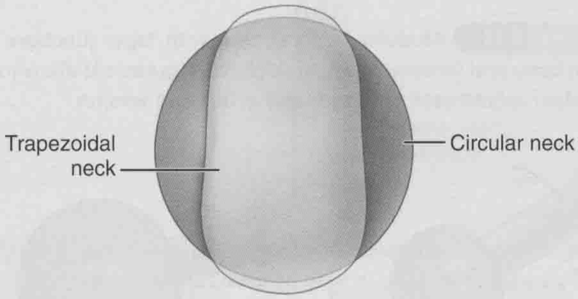


FIGURE 3-14 Cross-sectional comparison of circular and trapezoidal neck.

the use of larger head sizes than those traditionally used in the past.

In a range-of-motion simulation with digitized implants and virtual reality software, Barrack et al. found an improvement of 8 degrees of hip flexion when head size was increased from 28 to 32 mm. Range of motion was dramatically reduced by the use of a circular neck, especially when combined with a skirted modular head, which increases the diameter of the femoral neck (Fig. 3-13). A trapezoidal neck yielded greater range of motion without impingement than a circular one (Fig. 3-14). In an experimental range-of-motion model with head sizes larger than 32 mm, Burroughs et al. found that impingement between prosthetic components could be largely eliminated. When a head size larger than 38 mm was used, however, the only impingement was bone on bone and was dependent on bony anatomy and independent of head size. The ideal configuration of the prosthetic head and neck segment includes a trapezoidal neck and a larger diameter head without a skirt (see Fig. 3-18). In practical terms, the femoral head diameter is limited by the size of the acetabulum, regardless of the bearing materials used for the femoral head and acetabulum.

All total hip systems in current use achieve fixation of the femoral prosthesis with a metal stem that is inserted into the medullary canal. Much of the design innovation to increase prosthetic longevity has been directed toward improvement in fixation of the implant within the femoral canal. Many femoral stems have been in clinical use for variable periods since the 1990s. Recognition of the radiographic profile of a stem is often beneficial, however, in planning revision surgery. Readers are directed to previous editions of this text and other historical references for this information.

Femoral components are available in both cemented and cementless varieties.

CEMENTED FEMORAL COMPONENTS

With the introduction of the Charnley low-friction arthroplasty, acrylic cement became the standard for femoral component fixation. Advances in stem design and in the application of cement have dramatically improved the long-term survivorship of cemented stems. Despite these advances, the use of cement for femoral fixation has declined precipitously over the past decade and there has been little recent innovation in implant design.

Certain design features of cemented stems have become generally accepted. The stem should be fabricated of high-strength superalloy. Most designers favor cobalt-chrome alloy because its higher modulus of elasticity may reduce stresses within the proximal cement mantle. The cross section of the stem should have a broad medial border and preferably broader lateral border to load the proximal cement mantle in compression. Sharp edges produce local stress risers that may initiate fracture of the cement mantle and should be avoided. A collar aids in determining the depth of insertion and may diminish resorption of bone in the medial neck.

Mounting evidence suggests that failure of cemented stems is initiated at the prosthesis-cement interface with debonding and subsequent cement fracture. Various types of surface macrotexturing can improve the bond at this interface (Figs. 3-15 to 3-17). The practice of precoating the stem with polymethyl methacrylate (PMMA) has been associated with a higher than normal failure rate with some stem designs and has largely been abandoned. Noncircular shapes, such as a rounded rectangle or an ellipse, and surface irregularities, such as grooves or a longitudinal slot, also improve the rotational stability of the stem within the cement mantle (see Fig. 3-17).

There is concern that even with surface modifications the stem may not remain bonded to the cement. If debonding does occur, a stem with a roughened or textured surface generates more debris with motion than a stem with a smooth, polished surface. Higher rates of loosening and bone resorption were found with the use of an Exeter stem with a matte surface than with an identical stem with a polished surface. Similar findings have been reported when comparing the original polished Charnley stem with its subsequent matte-finish modification. For this reason, interest has been renewed in the use of polished stems for cemented applications. Ling recommended a design that is collarless, polished, and tapered in two planes (Fig. 3-18) to allow a small amount of subsidence and to maintain compressive stresses within the cement mantle.

Stems should be available in a variety of sizes (typically four to six) to allow the stem to occupy approximately 80% of the cross section of the medullary canal with an optimal cement mantle of approximately 4 mm proximally and 2 mm distally. Neutral stem placement within the canal lessens the chance of localized areas of thin cement mantle, which may