PROSTHETIC JOINT REPLACEMENT DESIGN
METHODS TO RESTORE KINEMATICS AND STABILITY AND PREVENT MATERIAL OVERLOAD

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ABSTRACT

To obtain improved outcome at a reduced cost, ***new*** total joint implant systems must allow the surgeon to restore joint mechanics without overloading prosthetic or skeletal materials. Normal joint mechanics depend on recreation of the joint's kinematic mechanism. In our work, mathematical modeling of joint surface shapes using average kinematic parameters results in duplication of the normal joint. However, a ***series*** of kinematically correct surfaces can be designed for any given joint. Surface and interface geometries can be varied to take into account the strengths and weaknesses of the prosthetic and biologic materials. The importance of surface asymmetry in ***stabilizing*** normal and prosthetic joints, and the processes used to create and analyze specific joint models are discussed.

INTRODUCTION

Many current total joint arthroplasty (TJA) prostheses do not accurately duplicate joint kinematics, making it impossible for the surgeon to restore normal motion and stability. Although restoration of useful motion can be completed without accurate restoration of normal mechanics, the chances for long term survival are reduced because the patient's bone and soft tissues must now resist increased forces (tension, torsion, shear and compressive stress concentrations) generated at the joint surfaces and particularly at the bone-prosthesis interfaces. Sometimes these forces are large enough to result in gross failure of the tissues, or the implant materials (McNamara et. al., 1994, Chiba et al. 1994, and Wright and Bartel, 1986). Often they are just large enough to significantly increase the generation of implant wear debris at the articular surfaces or the bone-prosthesis interfaces.

To deliver improved function and survivorship, designers must revisit *how* joints function, *how* surgical teams function, and *how* the disease process influences the reconstructive procedure. We are focusing on the *first* requirement in this paper.

HOW JOINTS FUNCTION

Understanding how joints move, how they provide stability under loading and how they are controlled is critical to restoration of function.

KINEMATICS: Several techniques have been used describe 3-space motion of human joints (Youm et al 1978, Youm and Flatt 1980, Chao and An 1982, Woltring et al 1985,Huskies et al 1985, Grood et al 1993). These kinematic descriptions are error prone (Woltring et al 1985, Hollerback et al 1994) and difficult to interpret. Recently, however, it has been documented that ***the motion of several joints is simple rotation about two or more fixed offset axes***.

JOINT	REFERENCE
ankle	Inman (1976) Singh et al (1992)
sub-talar	Inman (1976)
elbow	Weber & Weber (1836) Youm et al (1978) London (1980)
wrist	Sommers (1981) Moore et al (1993)

JOINT	REFERENCE
thumb carpometacarpal joint	Hollister et al (1992)
index metacarpophalangeal joint	Agee et al (1986) Brand and Hollister (1993)
forearm	Fick (1854) Hollister et al (1994)
knee tibio-femoral and patello-femoral joints	Hollister et al (1993)

These revolutes are **not** found within the traditional anatomic reference frames. They often do not intersect and are not perpendicular to each other or to the bone shafts.

MOTORS & MECHANICS: Tendons and muscles provide the motors. The mechanical advantage of the muscles and external forces are determined by their distance from and angle of application relative to the axes of rotation of the joints. Changing the location, nature or number of the joint's axes of rotation changes the spacial motion, the mechanics and the joint reaction forces for a given external load, effecting not only the resurfaced joint but the remaining joints in the extremity. Allowing more degrees of freedom in the prosthesis than are found in the natural joint usually results in a shortage of local motors (muscles) to control the joint.

MOTION: The SURFACE SHAPE of the bone-cartilage and meniscal components of joints control the spacial motion envelope. Meniscal structures function to increase stability, and yet provide more mobility than cartilage capped bone structures. Menisci are capable of compressive load transmission, function to distribute stresses over larger cartilaginous surface areas, and provide slightly greater stability than ligamentous structures. Their viscoelasticity protects underlying cartilage and bone under impact loading.

STABILIZERS: Ligaments and other soft tissues are stabilizers which help to keep the surfaces in close approximation so that the SHAPES of the joint components can maintain control of the motion envelope. **In positions which require high force transmission, bone surface shapes provide inherent stability to the joint** (Wolf's law).

LUBRICATION: Synovial fluid and membranes provide the lubrication and seal.

The kinematic mechanism can be used to predict the joint surface shapes and the joint motion envelope. **In our work, mathematical modeling of joint surface shapes using average kinematic parameters results in joint surface shapes which are <u>quite</u> close to those found in the normal joints.**

Once the kinematics of the joint have been and modeled and compared to the normal joint, regions of the bone surfaces which provide stability under load become apparent. The role of tissue stabilizers is also clarified**. Typically we have found that the sub-articular cortical bone surfaces are most congruent (and least flat) in positions of high loading**.

MATERIALS AND METHODS

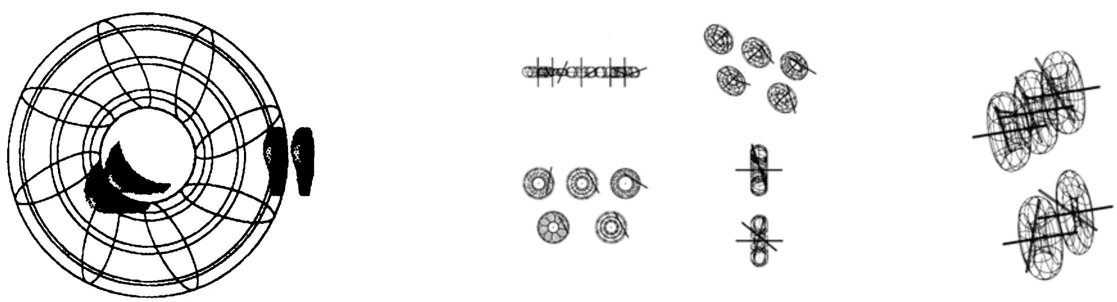
Computer aided design (CAD) and engineering analysis (CAE) software is used to create solid models of the implant surfaces (and subsequently of each component of the joint implant) whose surfaces are NURB surface patches bounded by curves. The software we used, SDRC IDEAS Master Series, integrates the solid modeling mathematics with finite element modeling (FEM) and analyses (FEA).

Articular surfaces for certain human joints can be modeled as a skewed torroid. The location of the revolutes with respect to the surface, and the bone anatomy will determine which portion of the torus is required to restore joint mechanics. At least two methods can be used to create skewed torroidal surfaces. The choice of the method depends on the CAD system. Both methods involve rotation of a curve around a circular arc. Only **sweeping** allows rotation of the actual offset conic. Swept surface modeling provides versatility. Different conics and NURB curves can be incorporated while maintaining the fixed revolutes of the joint, allowing enhancement of joint stability in specific locations. **Surfaces of revolution** must be created from geometry which lies in the same plane as the first revolute.

Implant surface shapes were analyzed to determine the effects of various parameters on joint mechanics and kinematics. Curvature (Atesian et al (1992)) , shape, range of motion (ROM) and stability comparisons between the normal joint and the prosthetic model were completed via computer and in cadaveric specimens, followed by worst case linear FEM/FEA.

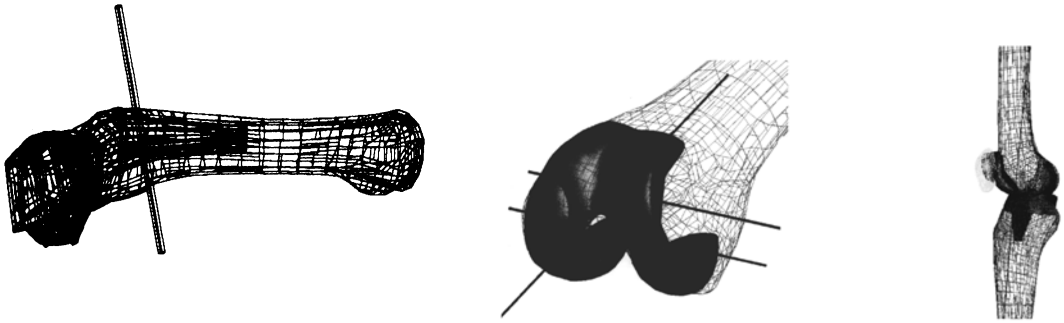
RESULTS

Changing the revolutes' orientation alters kinematics and the joint surface shapes. The amount of skew in the torus is determined by the degree of offset between the revolutes.



1a **1b** **1c**

Figure 1 - Shape Variation due to Offset Axes of Rotation. 1a -: Skewed torriod for the cmc joint created with swept surfaces. The location of the axes deter mines which portion of the surface is required for the joint. The saddle shapes on the left correspond to axes on opposite sides of the surfaces, while the ovoid shapes on the right correspond to axes on the same side of the surface. 1b: 4 Views of a series of torroids.. In each view, the top row of torroids show the shape changes associated with variation of the α angle, and the bottom row the more obvious changes associated with alterations of the β offset angles. 1c: Larger angled view of torroids in 1b



2a **2b** **2c**

Figure 2. Graphical shape comparisons. 2a: The thumb based cmc joint bones , implants and the joint's axes are shown. The flexion-extension axis and the abduction-adduction axis are also shown The FE axis is directed into the page (along the long axis of the trapezium). 2b: A left knee femoral implant model superimposed upon a wireframe knee model. 2c: A left femoral , tibial and patellar model superimposed upon the respective bones.

Graphical shape comparisons for a carpo-metacarpal joint (cmc) joint and a knee joint are shown. (***figure 2***). When comparing bones surfaces shapes, either physically or in the CAD environment, the regions of high bone congruency and inherent stability were always associated with peak load positions. The saddle shape of the patello-femoral joint and the medial transition region of the femur are good examples of this. In many joints, low load positions are not fully congruent. Note the varying levels of congruency associated with different joint positions in the cmc joint model (***figures 2a & 3***). The stability of the cmc joint in the highly loaded flexed position (***figure 2a***) is apparent. In the cmc joint model, high congruency was found in all neutral abduction-adduction orientations, and especially in the fully flexion positions associated with pinch and grip activities..

3a **3b**

Figure 3. Surface contact areas and congruency. 3a: The extended thumb is inherently less stable than the flexed thumb. The dorsal portion of the thumb surface matches the kinematic torroidal surface shape exactly. However, when the joint is rotated into full flexion, the shape changes to prevent continuous rotation about the FE axis , providing additional stability in high force activities. 3b: This series of illustrations show the limit of motion allowed by the cmc joint in flexion and extension. The top row illustrates extension motion, the center , flexion motion and the bottom, a portion of circumduction motion. More adduction than abduction is allowed with respect to the neutral position. Further Abduction-or adduction motion in these positions results in joint interference on one side and lift off on the opposing side. The joint is self- entering under high loads.

Flatter surfaces (associated with lower surface curvature and lower offset angles (α and β)) were less stable and had higher surface stresses than models with deep saddles. In the cmc joint prosthetic model, two arcs with constant radii were used to produce smooth surfaces (***figure 4***). However, the surfaces provide equivalent stability and higher congruency in high load positions compared to normal anatomy

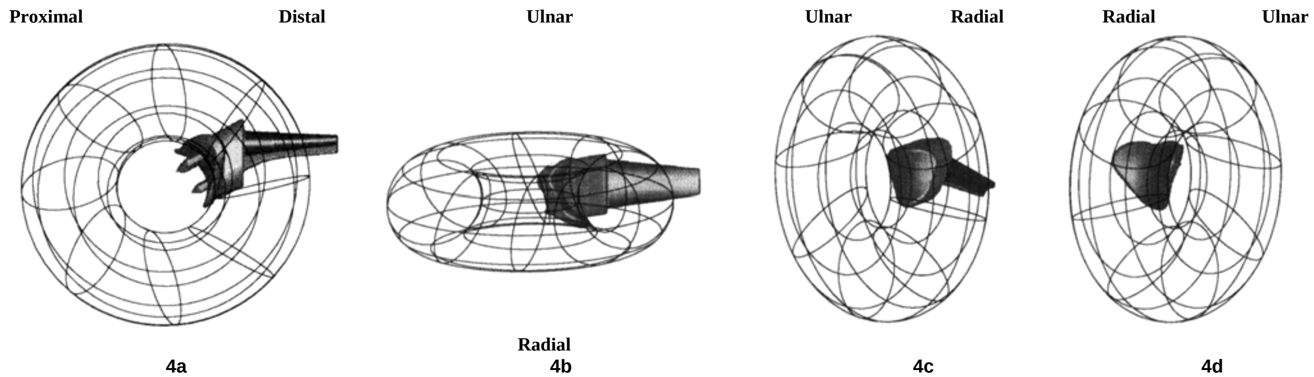


Figure 4. A left cmc joint implant superimposed on the torroidal swept surface used to create its articular surfaces. 4a: The dorsal surface of the metacarpal component stem (component on right) is positioned as it would be when properly aligned with the flat dorsal surface of the metacarpal bone. 4b: The implant as viewed when looking directly down at the dorsal surface of the metacarpal bone. Note the prominence of the trapezial surface on the radial side (lower side) . 4c: Metacarpal implant 's articular surface . 4d: Trapezial implant's articular surface. In this work, a thumb based cmc joint design has been created which meets the following objectives:

- The articular surface shapes and simplified surgical tools allow the surgeon to restore normal mechanics and kinematics of the joint.
- The prosthesis components transfer stresses to the cortical bone in a manner which prevents macromotion and hence minimizes wear debris at the bone-prosthesis interface.
- The prosthetic articular surface contact stresses and internal Von Mises stresses are well within acceptable performance levels for the materials used, minimizing the potential for articular surface wear debris and implant fatigue failure.

The worst case load in the cmc joint study, 2186 N (491 lb), corresponded to a 115.7 N (26 lb) pinch strength. The results are summarized in ***table 1***. Note the safety factors for each component. **By starting with a kinematic definition of a joint, it is possible to optimize the design to compensate for the inherent weakness of the polyethylene. It is possible to design highly congruent, kinematically accurate joint prostheses in which the polyethylene is not overloaded.**

Table 1 - Safety Factors Calculated from Worst Case FEA

Implant Component	Typical Worst Case FEA Results	Safety Factor Based on FEA
Metacarpal Metal FEA	8.96 MPa (13,000 psi)	8
Trapezial Metal FEA	148.00 MPa (21,500 psi)	3.5 - 82*
Metacarpal Polyethylene FEA	3.32 MPa (482 psi)	2** - 6

* The higher safety factor was calculated from results in a prior study in which rigid fixation of three pegs was presumed, and in which the peak stresses were 6.29 Mpa (913 psi). In this series, only one peg was rigidly fixed with nor surface support. It is not plausible that the pegs could be rigidly fixed without some sort of support at the saddle shaped interface. Thus the true safety factor for the Trapezial component is closer to 82 than to 3.5.

** The lower safety factor for polyethylene applies whenever the peak contact stresses rise above the yield limit of the material. It also compensates for errors associated with modeling the stress strain curve of this material as bilinear, and exacerbating conditions such as the use of thin cross sections of polyethylene and the potential for stress concentrations due to malpositioning. Stresses on the bone and/or bone cement were not analyzed in this study

The relationship between the kinematic surface shape, motion and stability is predictable for joints with fixed axes of rotation. Findings common to joints with two or more fixed offset axes of rotation:

1. **The motion about one or more axes is restricted by surface shape in positions where stability and/or high force transmission is required.**

2. **The surface shapes are oriented to resist the peak resultant forces in a manner which minimizes shear and torsion.**

3. **When the axes of rotation are not perpendicular to the bone shaft, the articular surface shapes are asymmetric .**

4. **The asymmetric surface shapes enhance the ability of the joint to resist dislocation and to transmit force.**

5. **The surfaces of revolution for two offset fixed axes can be used directly, indirectly, in whole, or in part to create implant articular surfaces.**

In the cmc joint and patella, greater saddle depth allows greater transmission of the shear forces, and minimizes the risk of translation or dislocation. The peak forces have direction vectors which point into the saddle shaped curves. The radial prominence of the cmc joint saddle nearly doubles the "effective" saddle depth making dislocation in this direction much less likely.

DISCUSSION

High force transmission is required of human joints. To optimize mechanical function many have evolved into shapes governed by the kinematic mechanism. Because bone tissue stronger in compression, it remodels in the direction of the highest loads, forming prominences. The surface shapes (bone prominences) contribute to joint stability under high loading by allowing compressive load transmission as opposed to shear and torsional force transmission. **The limited tensile strength of ligaments prevents their use as the primary stabilizer of the joint under high loads.** Motion about offset revolutes results in non-symmetric articular surface geometries for which fewer motors, ligaments or other tissues are required to maintain stability and position control.

These same surfaces (skewed torroids) can be used as articular surface geometry for TJA prostheses. Surface creation and modification techniques allow design variation to account for material properties and other considerations in prosthetic design while preserving the mechanics necessary for function. Creation of total joint implants with axes of rotation identical to those found in healthy normal joints, and with skewed torroidal surface components, will give surgeons a tool which **will** allow them to restore normal kinematics to the effected limb. *Moreover, the principles used to clarify the joint mechanics can be used to create new surgical tools and new surgical procedures which utilize bone grafts, cartilage grafts or synthetic bone/cartilage/mensical materials to restore the respective joint.*

References:

An, K., and Chao, E.: **Kinematic Analysis.** *In: Biomechanics of the Wrist Joint.* (An, K., Berger, R., and Cooney, W., Eds.), Springer - Verlag, New York, 1991, pp 29, 23-36.

Andrews, J., and Youm, Y.: **A Biomechanical Investigation of Wrist Kinematics**. *Journal of Biomechanics*, 1979, 12:83-93.

Ateshian, G., Rosenwasser, M., and Mow, V.: **Curvature Characteristics and Congruence of the Thumb Carpometacarpal Joint: Differences Between Female and Male Joints.** *Journal of Biomechanics, 1992*; 25:6:591-607**.**

Bartel, D., Bickell, V., and Wright, T.: **The Effect of Conformity, Thickness, and Material on Stresses in Ultra-high Molecular Weight Components for Total Joint Replacement**. *Journal of Bone and Joint Surgery*, 1986, **68a**: 1041-1051.

Brand, P., and Hollister, A. *Clinical Mechanics of the Hand, Second Edition.* , Mosby Year Book, Inc., St. Louis, 1992, pp 79-89, 106-27.

Buechel, F., and Pappas, M.: **The New Jersey Low-Contact-Stress Knee Replacement System: Biomechanical Rationale and Review of the First 123 Cemented Cases**. *Arch Orthop Trauma Surgery*, 1986, **105**:197-204.

Chao, E., and An, K: **Perspectives in Measurements and Modeling of Musceloskeletal Joint Dynamics**. In *Biomechanics: Principles and Applications* (R. Huskies et al., Eds.) Nijhoff, The Hague, 1-18, 1982.

Crosby, C., Wehbe, M., and Mawr, B.: **Hand Strength: Normative Values**. *Journal of Hand Surgery*, 1994,19A, 665-670.

Cooney, W., Chao, E., **Biomechanical Analysis of Static Forces in the Thumb During Hand Function**. *JBJS, 1977*; 59A:27-36**.**

Cooney, W., Lucca M, Chao, E., and Linscheid, R.: **The Kinesiology of the Thumb Trapeziometacarpal Joint**. *JBJS, 1981*; 63A:1371-81**.**

DeHeer: **Masters Thesis**, Purdue University, 1992.

Eaton, R., and Glickel, S.: **Trapeziometacarpal Osteoarthritis**, *Hand Clin*., 1987, **3**:455-471.

Fick, A.: *Die Gelenke Mit Sattleformigen Flachen. Zeitschrift fur Rationelle Medicin*. Heidelberg, Akademische Verlags-handlung von C.F. Winter, 1854, pp 314 - 321.

Fick, R.: *Handbuch der Anatomie und Mechanik der Gelenke unter Berucksichtigung der bewegenden Muskeln*. Part 1. Gustav Fischer Verlag, Jenna, 1904.

Fick, R.: *Handbuch der Anatomie und Mechanik der Gelenke unter Beruckisichtigung der bewegenden Muskeln*, 1908, Specielle Gelenk und Muskel Mechanik, Vol. II., Jena, Verlag von Gustav Fisher, 1911.

Giurantano, D. Personal Communications. Data has been submitted for publication, 1994.

Giurantano, D., and Hollister, A.: F**orce Analysis of the Thumb for a Five Link System**. ADM Vol. 120, 1991 Biomechanics Symposium ASME, 1991: 213-17.

Hollister, A., Buford, W, Myers, L., Giurantano, D., and Novick, A. : **The Axes of Rotation of the Thumb Carpometacarpal Joint**. *Journal of Orthopaedic Research 1992; 10:454-60.*

Hollister, A., Gellman, H., and Waters, R.: **The Relationship of the Interosseous Membrane to the Axis of Rotation of the Forearm**, *Clinical Orthopaedics and Related Research*, 1994, No. 298; 272-276.

Hollister, A., Jatana, S., Singh, A., Sullivan, W., and Lupichuk, A.: **The Axes of Rotation of the Knee**, *Clinical Orthopaedics and Related Research*, 1993, No. 290; 259-268 .

Huskies, R.: **Design, Fixation, and Stress Analysis of Permanent Orthopaedic Implants: The Hip Joint.**, *In: Functional Behavior of Orthopedic Biomaterials, Vol. II Applications*, **(**Ducheyne, P., and Hastings, G. Eds.), CRC Press, Inc., 1984a, pp. 125,148,155, 121-199.

Huskies, R.: **Principles and Methods of Solid Biomechanics,** *In: Functional Behavior of Orthopedic Biomaterials, Vol. I Fundamentals*, **(**Ducheyne, P., and Hastings, G. Eds.), CRC Press, Inc., 1984b, p. 88-97.

Huskies., R. and Blankevoort, L.: **The Relationship Between Knee Motion and Articular Surface Geometry**. *In: Biomechanics of Diarthrodial Joints* (Mow, V., Ratcliffe, A., and Woo, S., Eds.) Springer-Verlag, New York NY, 1990, pp 269-286.

Huskies R., Dijk, R., Lange, A., Woltring, H., and Rens, T.: **Kinematics of the Human Knee Joint**. *In: Biomechanics of Normal and Pathological Human Articulating Joints* (Berme, N., Engin,A., and L. Silva, C., Eds.) Nijoff, the Hague, 1985.

Imadeda, T.,An, K., W.,Cooney, W.,and Linscheid, R.: **Anatomy of Trapeziometacarpal Ligaments**. *Journal of Hand Surgery, 1993*; 18A:226-31**.**

Inman, V.: *The Joints of the Ankle*. Williams and Wilkins, New York, NY, 1976.

Inman, V., and Mann, R. : **Biomechanics of the Foot and Ankle**. *In: DuVrie's Surgery of the Foot*. The C.V. Mosby Co., St. Louis, MO, 1978.

Insall, J., Scott, W., and Ranawat, C.: **The Total Condylar Knee Prosthesis**. A Report of Two Hundred and Twenty Cases. *Journal of Bone abd Joint Surgery*, 1979, 61: 173-180.

Koebke, J.: *In: Advances in Anatomy, Embryology, and Cell Biology, Springer-Verlag* , Berlin, 1983, **80**:31-54.

Kuczynski, K.: **Carpometacarpal Joint of the Human Thumb**. *Journal of Anatomy*, 1974, **118**:119-126.

Lange, A., Huskies, R., Kauer, J., and Woltring, H.: **On the Application of Smoothing Procedure in the Kinematical Study of the Human Wrist Joint In-vitro**. *In: Biomechanics: Current Interdisciplinary Research* (Perren, S., and Schneider, E., Eds.) Martinus Nijhoff, The Hague, 1985, 303-308.

Lazar, G., and Schulter-Ellis, F.: **Intramedullary Structure of Human Metacarpals**. *Journal Of Hand Surgery*, 1980, 5:5: 477-481.

MacCoonaill, M: **Studies in the Mechanics of Ovial Joints : II. Displacements on the Articular Surfaces and the Significance of Saddle Joints**. *Ireland Journal of Medical Sciences*, 1946: **6**, 223-235.

McNamara, J., Collier, J., Mayor, M., and Jensen, R..: **A Comparison of Contact Pressures in Tibial and Patellar Total Knee Components Before and After Service in Vivo**. *Clinical Orthopaedics and Related Research.*1994: Jan.,Vol 299: 104-13.

Menon, J.: **The Problem of Trapeziometacarpal Degenerative Disease**. *Clinical Orthopaedics and Related Research*, 1983,**175**:155-165.

Meuli, C. : **Meuli Total Wrist Arthroplasty**. *Clinical Orthopaedics and Related Research*,1984, 187: 111.

Moore, J., Small, C., Hollister, A., and Giurantano, D.: **A Kinematic Technique for Describing Wrist Joint Motion**, Submitted for Publication, 1994.

Napier, J.: **The Form and Function of the Carpometacarpal Joint of the Thumb**, *Journal of Anatomy*,1955, **89**:362-369.

North, E., and Rutledge,W.: **The Trapezium-thumb Metacarpal Joint : The Relationship of Joint Shape and Degenerative Joint Disease**. *Hand*, 1983, 15:201-206.

Pappas, M., Makris, G., and Buechel, F.: **Contact Stresses in Metal-Plastic Total Knee Replacements: A Theoretical and Experimental Study**, Technical Report - Biomedical Engineering, Inc., 1986.

Pelligrini, V.: **Osteoarthritis of the Trapeziometacarpal Joint: The Pathophysiology of Articular Cartilage Degeneration. II Articular Wear Patterns in the Osteoarthritic Joint**. *Journal of Hand Surgery*, 1991, **16A**:975-982.

Rullkoetter, P., Anderson, D., and Hillberry, B.: **The Effects of Rotation and Sliding on the Stress State in UHMWPe Tibial Inserts**. *Transactions of the 40th Annual ORS,* New Orleans, LA., February, 1994, 801.

Singh, A., Starkweather, K., Jatana S., Hollister, A., and Lupichuk, A.: **Kinematics of the Ankle; A Hinge Axis Model**, *Foot and Ankle*, 1992; 13:439-446.

Sealy, F., and Smith, J.: *Advanced Mechanics of Materials, Chapter 14*, John Wiley and Sons, Inc., New York. NY, 1958.

Shigley, J.: *Mechanical Engineering Design, 3rd Ed.: Part I, Fundamentals of Design*. McGraw -Hill, Inc., New York, NY, 1972, pp. 26-73, 74-78, and 162-216.

Thomas, W.: **Uber die Aitiologie der Daumensattelgelenks arthrose und deren Behandlung durch eine spezielle Endoprosthese**. *Z. Orthop*., 1977, 115: 699-707.

Woltring, H., Huskies, R, Lange, A.,and Veldpaus, F.: **Finite Centroid and Helical Axis Estimation from Noisey Landmark Measurements in the Study of Human Joint Kinematics**, *Journal of Biomechanics*, 1985, 18:379-389.

Wright,T., and Bartel, D.: **The Problem of Surface Damage in Polyethylene Total Knee Components**. *Clinical Orthopaedics and Related Research*., 1986;73: 205.

Youm, Y., and Flatt, A.: **Kinematics of the Wrist**., *Clinical Orthopaedics and Related Research,* 1980; 149:21-32.

Youm, Y., McMurty, R., Flatt, A., and Gillespie, T.: **Kinematics of the Wrist, I. An Experimental Study of Radial-Ulnar Deviation and Flexion-Extension.** *Journal of Bone and Joint Surgery*, 1978, 60A:423.