

**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 revolute axes 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 quite close to those found in the normal joints.**

Once the kinematics of the joint have been 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.

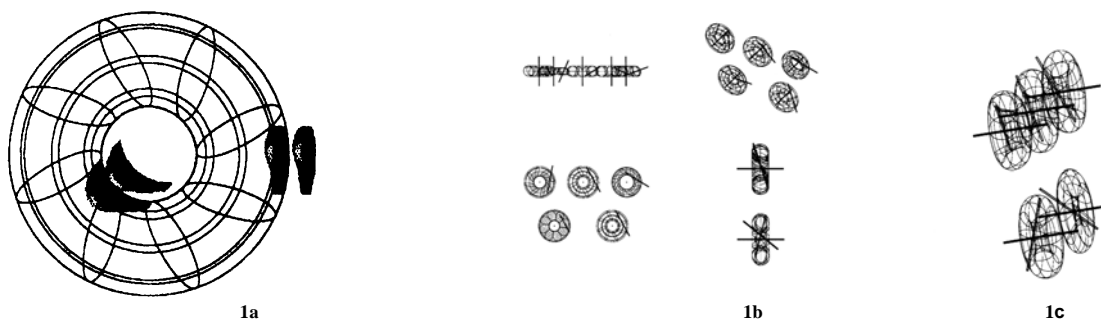


Figure 1 - Shape Variation due to Offset Axes of Rotation. 1a -: Skewed torroid for the cmc joint created with swept surfaces. The location of the axes determines which portion of the surface is required for the joint. The saddle shapes on the left correspond to axes on opposite sides of the surfaces, 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

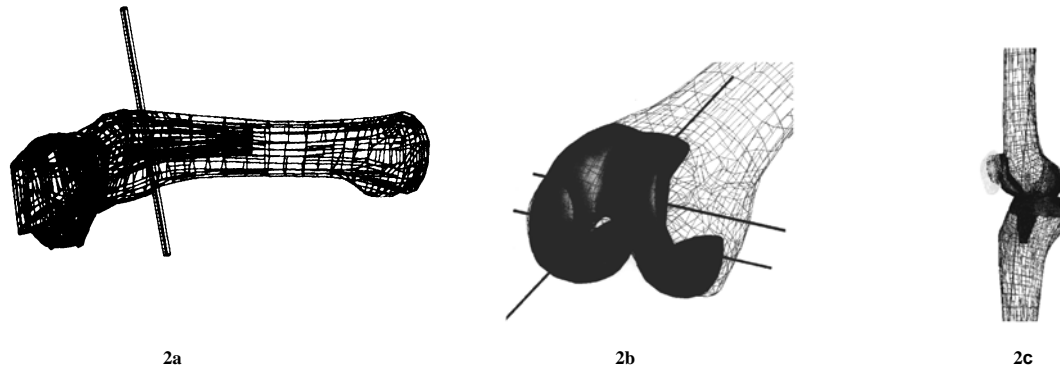


Figure 2. Graphical shape comparisons. 2a: The thumb based cmc joint bones , implants and the joint's axes are shown. The flexion-extension axis and the abduction-adduction axis are also shown. The FE axis is directed into the page (along the long axis of the trapezium). 2b: A left knee femoral implant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



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

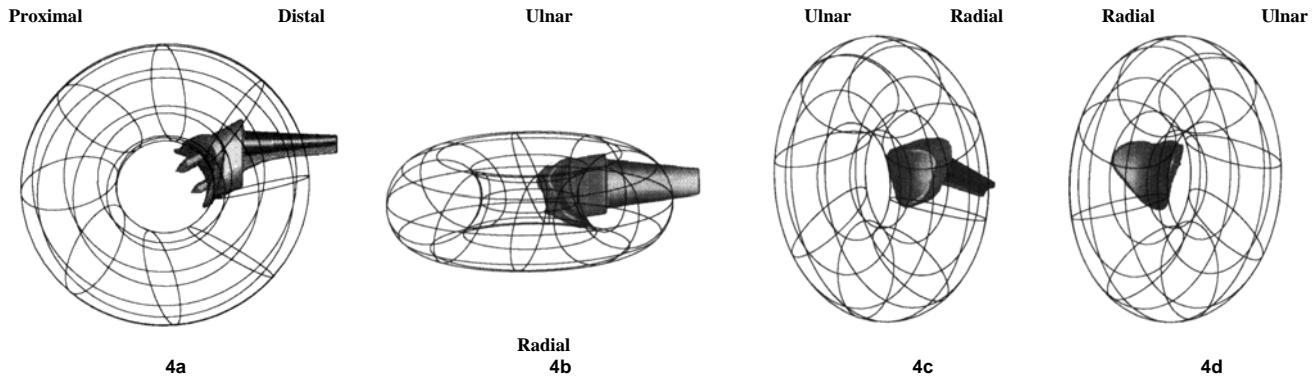


Figure 4. A left cmc joint implant superimposed on the torroidal swept surface used to create its articular surfaces. 4a: The dorsal surface of the metacarpal component stem (component on right) is positioned as it would be when properly aligned with the flat dorsal surface of the metacarpal bone. 4b: 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 no 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 revolves 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.*

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Prosthesis, prosthetic device or product Externally applied device used to replace wholly or partly an absent or deficient limb segment (6). Prosthetics Science and art of treating people by the use of prostheses (6). Prosthetist Person who has completed an approved course of education and training and is authorized by an appropriate national authority to design, measure and fit prostheses (6). vii. Standards for prosthetics and orthotics – part 1. standards. Prosthetist and orthotist Person who has completed an approved course of education and training and is authorized by an appropriate national authority. While an active prosthesis may help to restore the functional performance of an amputee, the requirements regarding the actuation unit as well as for the control system are high and the development becomes a challenging task. Regarding mechanical design and the actuation unit high force/torque delivery, high efficiency, low size and low weight are conflicting goals. The actuation principle and variable impedance actuators are discussed. Another design aspect found in above-knee prostheses concerns the kinematics of the knee joint. Since the human knee shows a rolling-sliding motion, the knee joint has an instantaneous center of rotation, which is defined by the centrode [47]. Recycling energy to restore impaired ankle function during human walking. PLoS ONE. Prosthetic joint replacement design methods to restore kinematics and stability while preventing material overload. Conference Paper. Full-text available. These surface shape variations can improve stability or congruency in different joint positions. A conic with an increasing radius about the trapezoidal axis from the center of the joint to the volar aspect of the trapezium is incorporated to recreate the saddle angle β . Closed section geometries are used during all modeling steps to create solid parts. The indication for a prosthetic joint replacement is a joint with advanced degenerative disease that is painful and may have associated stiffness and deformity, which is causing functional impairment. If there are no contraindications, a patient who is suffering from hemophilic arthropathy can look forward to the benefits provided by joint replacement surgery. Once it has been decided that a person with hemophilia will undergo a total joint replacement, educating the patient is crucial to a successful outcome. The following topics must be discussed prior to total joint replacement. 1. The person undergoing surgery needs to have a good understanding of hemophilia; how it has impacted the health of the joint and ultimately led to the need for joint replacement.