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

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Abstract - Using computer aided design tools, new total joint prostheses and surgical procedures have been designed to restore normal mechanics and thereby offer hope for improved patient outcome. 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. Current techniques often fail to restore normal joint function. Many replacement arthroplasties have experienced an unacceptable rate of loosening. Normal joint mechanics depend on recreation of the joint's kinematic mechanism. In our work average kinematic parameters are used to define mathematical models of the joint surfaces. Duplication of the normal joint is accomplished. 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.

New hand prosthetic design alternatives are discussed in detail. The importance of surface asymmetry in stabilizing normal and prosthetic joints, and the processes used to create and analyze specific joint models are also highlighted.

I. INTRODUCTION - REVIEW & THEORY

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 [1][2][3]. 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: Helical screw axis and Euler angle techniques [4][5][6][7][8] have been used to describe 3-space motion of most human joints. These processes are error prone [9]. Moreover, the resultant kinematic descriptions are difficult to utilize in the design process. Recently, however, it has been documented that ***the motion of several joints is simple rotation about two or more fixed offset axes***.

Table 1 - Joint Kinematics - References

JOINT	REFERENCE
ankle	Inman (1976, 1978) [10][11] Singh et al (1992) [12]
sub-talar	Inman (1976,1978) [10] [11]
elbow	Weber & Weber (1836) [13] Youm et al (1978) [14] London (1981) [15]
wrist	Summers (1984) [16] Moore et al (1993) [17]
thumb carpometacarpal joint	Hollister et al (1992) [18]
index metacarpo-phalangeal joint	Agee et al (1986) [19] Brand and Hollister (1992) [20]
forearm	Fick (1854) [21] Hollister et al (1994) [22]
knee tibio-femoral and patello-femoral joints	Hollister et al (1993) [23]

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 a joint controls the spacial motion envelope [21][24][25][26]. 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 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**.

II. 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 double precision non-uniform rational B-splines (NURB) surface patches bounded by curves. The software 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 [26][27]. 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. **Using this method, natural joint surface curvatures can often be incorporated with the kinematic definition to reproduce cartilaginous surface shapes with surface fit and curvature characteristics nearly identical to natural human specimens.** Surfaces of revolution must be created from non-variant 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 [28], shape, range of motion (ROM) and stability comparisons between the normal joint and the prosthetic model were completed both within the computer (via graphical, numerical and worst case linear FEM/FEA) and in cadaveric specimens.

III. RESULTS

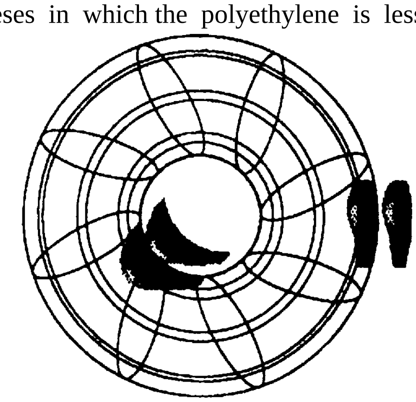
Changing the revolutes' orientation alters kinematics and the joint surface shapes. The amount of skew in the torus is determined by the degree of offset between the revolutes.

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 stability of the cmc joint in the highly loaded flexed position (***figure 2a***) is apparent.

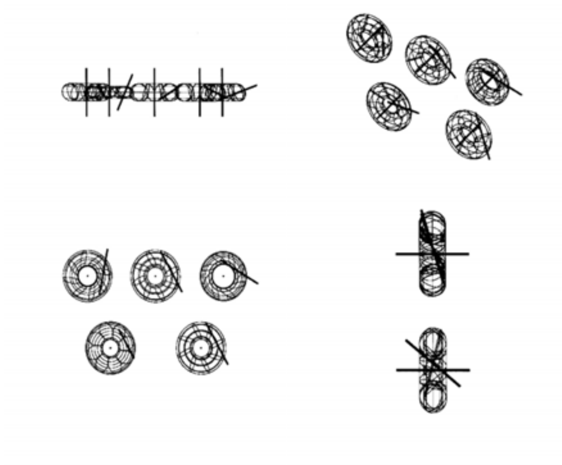
Flatter surfaces (associated with lower surface curvature and lower offset angles) were less stable and had higher surface stresses than models with deep saddles.

These results agree with those of MacCoonaill [29] and North and Rutlidge [30]. In the cmc joint prosthetic model, two arcs with constant radii were used to produce smooth surfaces (**figure 3**). However, the surfaces provide equivalent stability and higher congruency in high load positions compared to normal anatomy. A series of worst case finite element analyses is completed for each joint prosthesis. For the cmc joint the safety factors range from 2 to 82 as shown in table 2. The load in this series, 2186 N (491 lb), was derived from data published by Crosby [31] and Guirantano [32] [33], and corresponded to a 115.7 N (26 lb) pinch strength.

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 less subject to



1a



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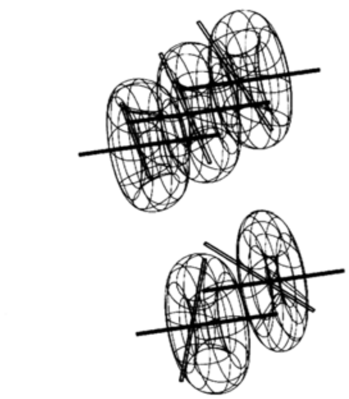
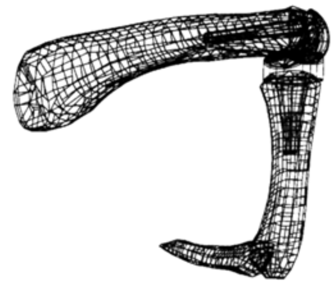
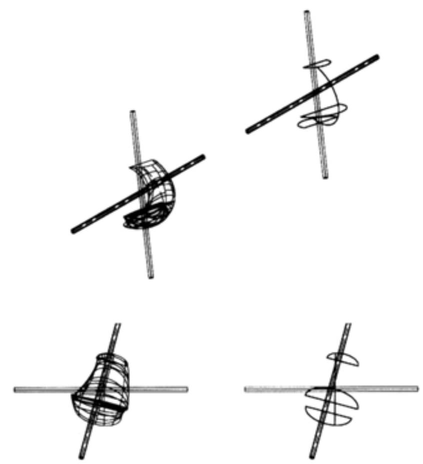


1c

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

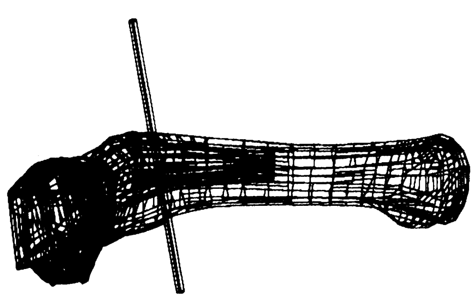


2b

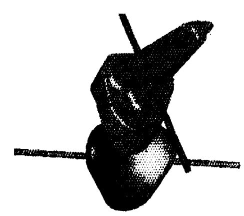


2c

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: PIP joint implant is shown (wireframe of surfaces) within a flexed middle finger proximal and interphalangeal bone. 2c: Two views of a left index finger MP joint swept surface, sweep creation geometries, and axes. Note that in this ovoid joint, the only way to keep the AB-AD axis constant is to vary the radii of the swept sections. Also note that without trimming or other modeling steps, the modified torroidal surface is quite close to the normal index finger MP shape.



2a



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- centering under high loads.

overload than in most less congruent (or “unconstrained” designs). Because the entire normal spacial motion envelope is accommodated by the articular design interface, forces at the “resurfaced” cortical bone-prosthesis interface should not differ significantly from normal. Addition of abnormal “constraint” is avoided with this design approach.

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. **Motion is restricted by surface shape in positions where stability is required.**
2. **Surface shapes minimize shear and torsion.**
3. **Articular surface shapes are asymmetric .**
4. **Asymmetry often helps to resist dislocation.**
5. **Surfaces of revolution for two offset fixed axes can be used to create implant articular surfaces.**

Table 2- Safety Factors Calculated from Simple Worst Case Linear 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. ** [34] [35] 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 [36][37], and exacerbating conditions such as the use of thin cross sections of polyethylene [38], and the potential for stress concentrations due to malpositioning. Stresses on the bone and/or bone cement were not analyzed in this study.

In saddle shaped joints like the cmc joint and the patella (in the knee joint), 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.

IV. DISCUSSION

Human joint structural geometries were created from the functional requirements of motion (kinematics) and stability (statics) for high force transmission. The surface and sub-surface shapes are the result of an evolutionary process which continually tries to optimize mechanical function. Because bone tissue is stronger in compression, it remodels in the direction of the highest stress, 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 most joints 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.

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