Predicting Hysteresis Loss in Hip Joint Implants

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Abstract : Wear is an important issue in hip implants. Excessive wear can lead to toxicity and other implant associated medical issues such as patient discomfort and decreased mobility. Since implant wear is result of contact between surfaces of femoral head and acetabulum implant, it is important to establish a model that can address implant surface roughness interaction.

A statistical contact model is developed for the interaction of femoral head and acetabulum implant in which surface roughness effects are included. The model accounts for the elastic-plastic interaction of the implant surface roughness. For this purpose femoral head and acetabulum implants are considered as macroscopically spherical surfaces containing micron-scale roughness. Approximate equations are obtained that relate the contact force to the mean surface separation explicitly. Closed form equations are obtained for hysteretic energy loss in implant using the approximate equations.

Keywords: Contact Mechanics – Roughness – Hip Implant – Wear – Energy Loss - Toxicity

1 Introduction

Hip joint serves as one of the most important load bearing joints in human body. Studies have shown that up to 5.5 times the bodyweight is tolerated by femur and pelvis during daily activities [1-3]. These include normal activities such as walking, going up or down a set of stairs, getting up or sitting down, carrying groceries or other loads. A hip joint provides, in addition to its load bearing ability, the needed mobility that includes extension, rotation, and flexion. Most importantly hip joints provide smooth articulation of limbs necessary for bi-pedal gait.

Hip joint malfunction may occur as a result of many factors. The most prevalent cause of hip joint surgical operation and hip joint replacement is osteoarthritis (OA). OA occurs when the cartilage fissuring is severe enough to a point where bone contact is initiated at the hip joint. OA is attributed to many causes [1] that include age, overuse, excessive loading, or flaw in the hip joint geometry referred to hip dysplasia. It is estimated that about 200,000 hip replacements occur in the United States due to hip joint OA. Other hip joint problems include osteolysis, avascular necrosis, neck fracture of femur [4-5-6]. The purpose of the present paper is not to address the causes of hip replacement, rather it is to address the performance of a hip joint after surgery.

Hip joint implant is designed to provide the same mobility and stability of the original functioning hip joint. Certainly, the design of hip joint implant needs to investigate all parameters such as wear, roughness, erosion, tribology, materials, and also many problems caused by surgical procedure including bone replacement. Some of these factors have been studied since about 50 years ago. About 50 years ago, McLaurin [7] investigated the manufacturing of hip At that time, the design encountered wear prostheses. problems because of metal on metal contact. Smith and Nephew [8] made an experimental model of hip joint using oxidized zirconium alloy technology in femoral head of hip joint to reduce wear and improve longevity in comparison with using ceramics for femoral head. In the last two decades, advances in imaging technology has allowed better preoperative data generation and improve preparation and planning of surgery [9-12]. A more recent work by Shapi et al. [13] allows preoperative measurement of the size of acetabular implant in total hip replacement.

Hiroyuki et al. [14] evaluated the effect of RF heating on hip joint implant during MRI examinations. They used two types of different implants in material and shapes. They found that the electrical characteristics of metallic implants have influence on RF heating. Maximum temperature was found to occur at the tip of the implants, location of large curvature. Zhang et al. [15] compared stress distributions between silicon nitride and cobalt-chromium-alloy in hip prostheses. The results related to stress distributions with the implanted silicon nitride hip resurfacing prostheses are very close to the corresponding stresses for health, intact femur bone. Scifert et al. [16] developed a new design to reduce the tendency of dislocation in Hip implants in patients. The authors claim that their proposed design increases stability of total hip joint and decreases by fifty percent stress distributions around impingement zone of polyethylene. Phillips et al. [17] used an elasto-plastic material model to show constitutive behavior of morsellised cortico-cancellous bone graft. Three 3D load scenarios related to walking, sitting, and standing were applied at the center of femoral head to check migration and rotation of the acetabular cup. Walking cause superior migration and rotation in abduction of the acetabular cup while sitting down and standing up cause posterior migration and rotation of the acetabular cup. Jonathon et al. [18] investigated dangerous effects of metal release from hip prostheses on patients.

Metal-on-metal hip prostheses failed in some patients due to the release of metal debris resulting in revision surgery.

Symptoms such as neurological impairment, cardiomyopathy, and hypothyroidism were reported in their study. Steens et al. [19] showed the effect of ceramic-on- ceramic toxicity in the blood can lead to impairment of hearing, sight, numbness in feet, and dermatitis in head and neck. Tower [20] also showed that the dangerous effects of metal debris in human blood pain such as onset of anxiety, major depression, tinnitus, high frequency hearing. loss, peripheral neuropathy, and cognitive decline. Alan and Swarts [21] investigated the effect of modularity on tapered cone of Margron hip prosthesis. Their study found that increased modularity can cause corrosion and crack, debris of particles, and metal ion generation. Brodner et al. [22] investigated the levels of serum cobalt in patients before and after implantation of noncemented total hip arthroplasties. As a result, they show that the metal-on-metal prostheses produce detectable levels of serum cobalt in comparison with the ceramic-polyethylene prostheses as metal-on-metal prostheses generate some systemic release of cobalt.

This paper develops a contact mechanics model of hip joint taking into account the effect the surface finish property and surface roughness geometry of the implant. An elasticplastic model of the spheres in contact representing the femoral and acetabular implants is developed.

The specific contribution of this paper includes:

- Inclusion of implant surface roughness in hip implant contact model
- Approximate equations relating the contact force to minimum mean plane separation in an explicit form
- Energy loss per cycle that include macro and micro geometry of the implant surfaces
- Characterization of hip implant natural contact frequency and contact damping

The results agree with the recent issues with hip implant failures when metal-on-metal is employed. The model presented is, therefore, a necessary first step in the prediction of possible wear in hip implants and issues related to wear borne toxicity in implant recipients.

2 Hip Contact Model:

The schematic diagram of a hip joint is shown in Fig 1. Figure 1 shows that force transfer to hip gives rise to contact force between the femoral head and acétabulés, whose shapes are approximated using spheres. Let R_1 and R_2 be the radii of curvature of the femoral head and acetabulum, respectively. Figure 2 details the contact between two spheres of radii R_1 and R_2 . When roughness of the surfaces is incorporated into the contact model, it is expected that the load-carrying zone be defined by a minimum separation with symmetrically distributed pressure about the minimum separation. Since the number of contact points and their respective pressure depend on the mean surface separation of the two spheres, it is necessary to develop the expression for mean separation as a function of minimum separation and the geometries of the two spheres. In contact of femoral head with acetabulum, we confront a conformal contact. This is represented by the sphere contact in Fig. 1.



Figure 1. Schematic depiction of contact force in hip joint



Figure 2. Spheres in internal contact

The schematic drawing of the mean spheres of the femoral head and the acetabulum surface in Fig. 2 shows that for a mean surface separation h_{0} , the offset between sphere centers, δ , can be expressed in terms of h_{0} .

$$\delta = R_2 - R_1 - h_0 \tag{1}$$
$$R_1 + h = x \tag{2}$$

Where, the triangle shown in Fig 2, clearly shows that the mean plane separation, h, can be found in terms of minimum separation, h_0 , radii of the two spheres, R_1 and R_2 , and the angular location measured with respect to the inner (smaller) sphere.

$$x = -\delta\cos\theta \pm \sqrt{\delta^2\cos^2\theta + R_2^2 - \delta^2}$$
(3)

An acceptable solution in Eq. (3) must yield a positive x. Therefore,

$$x = -\delta \cos\theta + \sqrt{\delta^2 \sin^2\theta + R_2^2} - \delta^2 \tag{4}$$

Substitute for x in terms of R_1 and h and solve the resulting equation for h, the separation at location θ . We find from eq. (2) and (4),

$$h = R_2 \left(\left(-\frac{\delta}{R_2} \right) \cos\theta + \sqrt{1 - \left(\frac{\delta}{R_2}\right)^2 \sin^2\theta} \right) - R_1 \qquad (5)$$

Substitute for δ from eq. (1) to find

$$h = R_2 \left(\left(-\frac{R_2 - R_1 - h_0}{R_2} \right) \cos\theta + \sqrt{1 - \left(\frac{R_2 - R_1 - h_0}{R_2} \right)^2 \sin^2\theta} \right) - R_1$$
(6)

Since the mean surface separation is defined, we can proceed to derive the contact force per unit nominal area due to elasticplastic interaction of the roughness of the femoral and acetabulum surfaces. The contact force per unit nominal area can be expressed as follows [23],

$$P(h) = P_{ec}(h) + P_{p}(h)$$
(7)

Where, $P_{e}(h)$ is the elastic force per unit nominal area given by the following equation,[23],

$$P_{e}(h) = C\left(\int_{h}^{\infty} (s-h)^{\frac{3}{2}} e^{-\frac{s^{*}}{2}} ds - \int_{W_{c}+h}^{\infty} (s-h)^{\frac{3}{2}} e^{-\frac{s^{*}}{2}} ds\right)$$
(8)

Where,

$$C = \frac{4}{3\sqrt{2\pi}} E\eta \beta^{\frac{1}{2}} \sigma^2 \tag{9}$$

s and h are both dimensionless. s is the ratio of an asperity height over the standard deviation of asperity summit distribution, σ , and h is the ratio of the mean surface separation over σ . When the surfaces are pressed together, there may be locations within the contact zone where asperity interference results in onset of plastic deformation. Greenwood and Williamson [24] defines asperity critical interference to be the onset of plastic deformation. In following the CEB model [25], the authors employed the definition of the critical interference to formulate the elasticplastic model of contact. In eq. (8) w_c represents the dimensionless critical interference. Greenwood and Williamson [24] defines plasticity index and critical interference for a surface as follows:

$$\psi = \frac{E}{H} \sqrt{\frac{\sigma}{R}} \qquad \omega_c = \left(\frac{H}{E}\right)^2 R$$

Where, *R* is the average asperity summit radius of curvature, *E* is the equivalent modulus and *H* is the hardness of the softer material. Letting $w_c = \frac{\omega_c}{\sigma}$ be the dimensionless critical interference, the plasticity index, ψ , is related to the w_c as follows

$$\psi = \frac{1}{\sqrt{w_c}} \tag{10}$$

Equation (8) uses a constant *C* and the dimensionless force expression in integral form for the elastic part of the surface interaction. *C* is defined as given by eq. (9), in which *E* is the reduced modulus of elasticity of the two surfaces, β is the dimensionless equivalent average asperity radius of curvature. The reduced modulus of elasticity is derived from the properties of the material used in the implant. It is given by the following equation

$$\frac{1}{E} = \frac{1 - v_1^2}{E_1} + \frac{1 - v_2^2}{E_2} \tag{11}$$

Where, E_1 and ν_1 are the modulus of elasticity and Poisson ratio of the femoral implant material and E_2 and ν_2 are those of the acetabulum implant. The equivalent asperity radius is found using

$$\frac{1}{\beta} = \frac{1}{\beta_1} + \frac{1}{\beta_2} \tag{12}$$

Where, β_1 and β_2 are the average asperity radius of curvature of the femoral and acetabulum implants, respectively. η is the asperity density per unit area. The force per unit normal area due to plastic interaction in an elastic-plastic contact is [58, 60].

$$P_{p}(h) = C\left(\frac{3}{2}\sqrt{w_{c}}\int_{w_{c}+h}^{\infty}[2(s-h)-w_{c}]e^{-\frac{s^{2}}{2}}ds\right) (13)$$

To obtain the contact force along a particular direction, one must sum the force components along that direction due to infinitesimal contact forces that occur over an infinitesimal area. Sum of the contact infinitesimal contact forces along the line of symmetry will be result in the total contact force, whereas those along the normal to the line symmetry vanish. Sum the force components parallel to the radial line of symmetry with respect to the nominal contact area to find

$$F = \int_0^{2\pi} \int_{\frac{\pi}{2}}^{\frac{\pi}{2}} P(h) R^2 \sin\theta \, \cos\theta \, d\theta d\phi \tag{14}$$

Where, R is the equivalent macro radius of curvature of the spheres representing femoral and acetabulum implants

$$\frac{1}{R} = \frac{1}{R_1} + \frac{1}{R_2}$$
(15)

The integral in eq (14) can be reduced to the following

$$F_n = 4\pi\sigma^2 \int_0^{\overline{2}} P(h) R^2 \sin\theta \,\cos\theta \,d\theta \tag{16}$$

It will prove beneficial to express eq (16) as an explicit function of the minimum separation, h_0 . Since *h* is a function of integration variable Θ , eq (6), the integral in eq (16) can only be found numerically. As a result, we set out to find an approximate relation between contact force *F* and the minimum separation h_0 . It will be shown in the next section that the contact force may be estimated using a function of the form $\aleph e^{-ch_0^{b_1}}$.

2.1 Dependence of Coefficients on Hip Radii

In this section acetabulum and femoral radii are used as parameters in the approximate expression relating contact force to minimum mean surface separation. It can be shown that the approximate equation is of the following form

$$F_{na}(h_0, R_1, R_2) = \aleph(\psi, R_1, R_2) e^{-ch_0^{1/2}}$$
(17)

The values are generated for various femoral head radii, ranging from 5 mm to 25 mm. Where, the coefficients \aleph and c are expected to depend on the geometry of the hip and the plasticity index.

$$F_n(h_0, R_1, R_2) = 4\pi \sigma^2 \int_0^{\frac{\pi}{2}} P(h) R^2 \sin\theta \, \cos\theta \, d\theta \qquad (18)$$

In obtaining the approximate equation, the femoral radius is varied, and the acetabulum radius is assumed to be 0.2 mm larger than the femoral radius. Femoral radius is varied from 5 mm to 25 mm [61] while in each case acetabulum radius is kept 0.2 mm larger.

2.2 Dependence of Coefficients on Plasticity Index ψ

In this section, approximate functional relationships between the coefficients and plasticity index are established

for plasticity index ranging 0.3 to 1.3. Keep in mind that for surfaces characterized by $\psi < 0.6$ the surface is considered predominantly elastic, while for $0.6 < \psi < 1$ the surface is viewed as elastic-plastic.

$$\aleph(\psi, R_1, R_2) = a(\psi)R_1^2 + b(\psi)R_1$$
(21)

$$a(\psi) = a_3\psi^3 + a_2\psi^2 + a_1\psi + a_0 \tag{22}$$

$$a_3 = -6.179 \times 10^{-5}, a_2 = 1.65 \times 10^{-4},$$

$$a_1 = -1.237 \times 10^{-4}, a_0 = 7.132 \times 10^{-3}$$
(23)
Likewise, the fitted function for b is

$$b(\psi) = b_3 \psi^3 + b_2 \psi^2 + b_1 \psi + b_0$$
(24)

with coefficients

$$b_3 = -0.012313, b_2 = 0.032885,$$

$$b_1 = -0.024663, b_0 = 0.010218$$
 (25)
The function $c(w)$ is defined as follows

$$c(\psi) = c_0 + c_1 \psi + c_2 \psi^2 + c_3 \psi^3$$
(26)

Where,

$$c_3 = -0.118687, c_2 = -0.069481,$$

 $c_1 = 0.513741, c_0 = 1.7366$ (27)

Finally, plasticity function with low percent error for a, b, and c is

$$F_{na}(h_0, R, \psi) = [a(\psi)R_1^2 + b(\psi)R_1]e^{-c(\psi)h_0^{1/2}}$$
(28)

The max error between the approximate and original elasticplastic contact force is less than 5% over the entire range of parameters considered.

2.3 **Energy Loss in Hip Implant**

The contact between femoral and acetabulum implant surfaces consists of asperities experiencing elastic and plastic deformation. A close look at the loading and unloading process reveals that both energy loss and elastic recovery are involved in the process. During the increase in contact load both elastic and plastic deformations can occur at asperity deformation level. However, during unloading asperities undergo only elastic recovery. Therefore, the load and unload process will follow different paths, resulting in hysteresis type energy loss in the hip joint contact.

We can employ the approximate equations for elasticplastic contact and purely elastic contact to represent the loading and unloading process mathematically. The force during loading is denoted $F_{nL} = \alpha_{1L} e^{\alpha_{2L}h_0^{\alpha_{3L}}}$ and that during unloading, $F_{nU} = \alpha_{1U} e^{\alpha_{2U}h_0^{\alpha_{3U}}}$. Based on the results of the previous section, the respective coefficients of contact force during load and unload are as follows:

$$\alpha_{1L} = a_L(\psi)R_1^2 + b_L(\psi)R_1$$
(30)

$$\alpha_{2L} = -c_L(\psi) \tag{31}$$

$$a_{3L} = a_{3U} = 1.2 \tag{32}$$

$$a_{41} = a_{41}(\psi)R^{2} + h_{41}(\psi)R. \tag{33}$$

$$a_{1U} - a_{U}(\psi) \kappa_{1} + b_{U}(\psi) \kappa_{1}$$
(33)

$$\alpha_{2U} = -c_U(\psi) \tag{34}$$

To study energy loss and storage in a hip joint, we consider an equilibrium contact force. For example this may correspond to an individual standing still and a contact force equal to the equilibrium force exists between femoral head and acetabulum. The equilibrium contact force is associated with an equilibrium minimum mean plane separation, h_0 . A disturbance from equilibrium is denoted x. Therefore, to

study the behavior of the contact near an equilibrium state, we can use the contact force equations above. Depending on the nature of the disturbance, the load may increase from equilibrium or decrease from it. If the load is increasing from equilibrium then both elastic and plastic contacts must be included in the calculation of contact force. If the load is decreasing from the equilibrium state, then only elastic contacts contribute, since this is a load recovery process. The following expressions will be adequate to account for either load change scenarios.

$$F_{nL}(h_0, x) = \alpha_{1L} e^{\alpha_{2L}(h_0 - x)^{\alpha_{3L}}}$$
(35)

$$F_{nU}(h_0, x) = \alpha_{1U} e^{\alpha_{2U}(h_0 - x)^{\alpha_{3U}}}$$
(36)

Here F_{nL} denotes the normal contact load due to both elastic and plastic interaction of surface roughness, and F_{nU} is the normal contact force due to only elastic interaction of the roughness. When the disturbance is small, the above force equations can be written in linear form using truncated Taylor series expansion of F_{nL} and F_{nU} about the equilibrium minimum separation.

$$FL_{nL}(h_{0},x) = -\left(\frac{h_{0}^{\alpha_{3L}-1}\alpha_{1L}\alpha_{2L}\alpha_{3L}e^{h_{0}^{\alpha_{3L}}\alpha_{2L}}}{1!}\right)x + \alpha_{1L}e^{\alpha_{2L}h_{0}^{\alpha_{3L}}}$$

$$FL_{nU}(h_{0},x) = -\left(\frac{h_{0}^{\alpha_{3U}-1}\alpha_{1U}\alpha_{2U}\alpha_{3U}e^{h_{0}^{\alpha_{3U}}\alpha_{2U}}}{1!}\right)x + \alpha_{1U}e^{\alpha_{2U}h_{0}^{\alpha_{3U}}} x + \alpha_{1U}e^{\alpha_{2U}h_{0}^{\alpha_{3U}}}$$
(37)

(38)

Figure 4 illustrated the contact forces along with their linear estimates about an equilibrium position for a relatively high plasticity index. The area between the load and unload forces represents energy loss per cycle. Figure 5, shows a similar force history corresponding to a lower value of the plasticity index. As expected the area between the load and unload phases are reduced to zero for a plasticity index of 0.5, since it corresponds to elastic behavior of contact. It is a simple task to estimate the energy loss per cycle.



Figure 4. The schematic of load – unload phases in high plastic zone



Figure 5. The schematic of load – unload phases in low plastic zone

We can perform integration of force over displacement in the load and unload phases and obtain the energy loss in a single cycle. For amplitude of oscillation of x_a from equilibrium, we can express the energy loss per cycle as follows.

$$E = \int_{-x_a}^{x_a} F_{nL} \, dx - \int_{-x_a}^{x_a} F_{nU} \, dx \tag{39}$$

That can be simplified by using the linear approximation of each load function in eqs (37) and (38). We find

$$E_{L} = 2C x_{a} \left(\alpha_{1L} e^{\alpha_{2L} h_{0}^{\alpha_{3L}}} - \alpha_{1U} e^{\alpha_{2U} h_{0}^{\alpha_{3U}}} \right)$$
(40)

 E_L is the energy loss per cycle. The energy per cycle can be expressed in dimensionless form by dividing eq. (40) by C x_a . So the dimensionless energy loss per cycle is

$$E_{L} = 2 \left(\alpha_{1L} e^{\alpha_{2L} h_{0}^{\alpha_{3L}}} - \alpha_{1U} e^{\alpha_{2U} h_{0}^{\alpha_{3U}}} \right)$$
(41)

Figure 6 illustrates dimensionless energy per cycle and plasticity index as functions of dimensionless critical interference. When critical interference is low (high plasticity index), the interference enters the plastic regime for less contact load. Therefore, energy loss per cycle is higher for low critical interference. As critical interference increase, the number of asperities experiencing plastic interference decrease, thereby, reducing the energy per cycle. This is clearly shown to be the case in Fig. 6.

A similar plot, which directly relates energy loss per cycle to surface roughness, is shown in Fig. 7. In this case the abscissa represents the dimensionless average radius of curvature. It is observed that as dimensionless asperity summit radius of curvature is increased (surface is made more smooth) the energy loss per cycle decreases.



Figure 6. Dimensionless energy loss per and surface plasticity index versus critical asperity interference.



Figure 7. Dimensionless energy loss per cycle and surface plasticity index versus dimensionless average asperity summit radius of curvature

Recent legal litigation regarding the use of similar implant material shed light on the difficulty faced in using similar material in hip implant replacement. Cobalt-Cobalt implant was alleged to result in excessive degradation of implant material, generating unacceptable amount of wear debris to the level of presenting toxicity in the patient. Based on the present study, use of Cobalt-Cobalt implant would require a very high level of surface finish to reduce plastic deformation. Consider, for example, the equation for plasticity index given by Greenwood and Williamson [59]

$$\psi = \frac{E}{H} \sqrt{\frac{\sigma}{R}}$$

Гał	ole	1.	Material	and	surface	pro	perties	used
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Parts	Materials	Y M	'oung's Iodulus	Poisson's Ratio	Geometry (mm)					
Femoral Head	Cobalt- Chromium- Alloy	200 Gpa		0.3	Sphere Radius 5-25 mm					
Acetabu lum	Polyethylen e	0.64 Gpa		0.4	Sphere Radius 6-26 mm					
Surface Properties										
	property		values							
η (asper	rity/area density)		$11 \times 10^{10} \frac{1}{m^2}$							
σ Star	ndard deviation		$0.5 \times 10^{-6} m$							
β Mean d	asperity rad (R/σ))	200							
E equiv. e	elasticity modulus	5	7.572 × 10 [#] Gpa							

For Cobalt-on-Cobalt implant we can use the properties in Table 1 to find E = 62 GPa and H = 700 MPa. For average dimensionless asperity summit radius of curvature $\beta = R/\sigma =$ 4600, we obtain a surface plasticity index of $\psi = 1.3$ which puts deformation in the plastic range. In fact to ensure that contact is in the elastic range, the surface finish must be enhanced to a degree that would result in $\beta = 21,600$, an unrealistically high number. This result is consistent with the latest news regarding the failure of many implants involving metal-on-metal material.

Figure 7 corresponds to typical ranges when dissimilar materials are used in hip implant. For Cobalt-Polyethylene, we use E = 0.75 GPa, and H = 120 MPa (Table 1) for the Polyethylene, the softer material, and $\beta = 25$, the surface plasticity index $\psi = 0.953$, putting the contact in the elastic-plastic range, whereas for $\beta = 112$, the plasticity index reduces to $\psi = 0.586$, yielding elastic contact. The plot in Fig. 7 well represents this range. The advantage of using dissimilar material is clearly shown in the above discussion. It is easy to obtain surface finish that would yield an elastic interaction at the contact of femoral head and acetabulum implant if one employs dissimilar material, while using similar material, such as Cobalt-Cobalt, is quite problematic since to guarantee elastic contact the surface finish requirement are not attainable.

3 Closing Remarks

This paper has developed an elastic-plastic contact model of hip joint implant. The model treats femoral and acetabulum implants as spherical solids in internal conformal contact and accounts for the roughness effects of both surfaces. An equation relating force to minimum mean surface separation was derived using statistical integral over contact region of effective interaction. Approximate equation describing force explicitly in terms of minimum separation was obtained and used to find closed-form equation for contact energy loss per cycle. It is shown that energy loss per cycle varies with plasticity index of the surface of the weaker implant. For an assumed lump mass representation at contact of implant, the utility of the approximate equations were exemplified by deriving expressions for contact natural frequency and damping ratio.

The specific contribution of this paper includes:

- -Inclusion of implant surface roughness in hip implant contact model
- -Explicit function for hip implant for a considerable range of hip implant joint sizes
- -Energy loss per cycle related to macro and micro geometry of the implant surfaces
 - Closed-form equation for hysteristic energy loss per cycle was obtained using load/unload process at the hip implant surfaces. The energy per cycle was related explicitly to the material properties and surface statistics of the implant.

Simulation of load/unload process on hip implant using similar material, Cobalt-Cobalt, and dissimilar material, Cobalt-Polyethylene, for the femoral head and acetabulum showed that:

- -Similar implant material was more prone to plastic deformation, thereby suggesting the increased possibility of wear. Cobalt-Cobalt contact required an unreasonably high surface finish to minimize plastic energy loss. Such high requirement of surface finish is impractical and even if possible would be highly costly.
- -Dissimilar implant material was shown to be superior in that it is easier to guarantee elastic contact so that the plastic energy loss is minimized for a practical range of surface roughness.

The above result is consistent with recent issues related to the use of similar material in hip implant. Recent litigation won by an implant patient in California against Johnson and Johnson related to failed implant due to the generation of excessive wear and the resulting toxicity. According to the news, the reason was primarily due to the use of metal-onmetal in the hip implant. Many more lawsuits relating to the metal-on-metal contact in hip implants against Johnson and Johnson are being submitted to the courts.

The potential usefulness of the results on estimation of hip implant contact frequency and damping can involve the issue of potential vibration and noise generation. These not only can result in accelerated fatigue wear of implant surfaces but also can relate to implant recipient's comfort level.

4 Nomenclature

- *C* a surface constant
- *E* equivalent modulus of elasticity of the two surfaces
- E_1 modulus of elasticity of the femoral implant
- E_2 modulus of elasticity of the acetabulum implant
- E_L energy loss per cycle
- *F* total contact force
- F_{nL} normal contact load due to elastic-plastic interaction of roughness
- F_{nU} normal contact load due to elastic interaction of roughness
- *h* mean plan separation
- h_0 minimum separation
- m_0 mass of femoral head
- *Pe(h)* elastic force per unit nominal area
- Pp(h) plastic force per unit nominal area
- P(h) contact force per unit nominal area
- R_1 radius of femoral head
- *R*₂ radius of acetabulum
- *s* ratio of an asperity height over the standard deviation *c* critical interference
- *x* a disturbance from equilibrium
- x_a amplitude of oscillation
- α_{1L} , α_{2L} , and α_{3L} coefficients in the approximate function for loading phase
- α_{1U} , α_{2U} , and α_{3U} coefficients in the approximate function for unloading phase
- β dimensionless equivalent average asperity radius of curvature
- β_1 average asperity radius of curvature of the femoral implant
- β_2 average asperity radius of curvature of the acetabulum implant
- δ offset between sphere centers
- ζ damping ration
- η the asperity density per unit area
- θ angular location
- v_1 Poisson ratio of the femoral implant
- *v*₂ Poisson ratio of the acetabulum implant
- σ standard deviation of asperity height
- ψ plasticity index
- ω_n natural frequency

5 References:

[1] Hodge, W. A., Fijan, R. S., Carlson, K. L., Burgess, R. G., Harris, W. H., and Mann, R. W. Contact Pressures in the Human Hip Joint Measured in Vivo. Proc Natl Acad Sci USA 83,. 2879-83,. 1986.

[2] Bergmann, G., Deuretzbacher, G., Heller, M., Graichen, F., Rohlmann, A., Strauss, J., and Duda, G. N. Hip Contact Forces and Gait Patterns from Routine Activities. J Biomech 34,. 859-71,.2001. [3] Bergmann, G., Hip98: Data Collection of Hip Joint Loading on CD-Rom. Free University and Humboldt University, Berlin, 1998.

[4] Pramanik, S., Agarwal, A., K., and Rai, K., N. Chronology of Total Hip Joint Replacement and Materials Development. Trends Biomater. Artif. Organs,19,.1,.15-26,.2005.

[5] Charbonnier, C., Schmid, J., Kolo-Christophe, F., Magnenat-Thalmann, N., Becker, C., and Hoffmeyer, P. Virtual Hip Joint: from Computer Graphics to Computer-Assisted Diagnosis. Eurographics, 2009.

[6] Montecucchi P. C., "Total Anatomic Hip Prosthesis. Montegen,540 Beverly Court, Suite1- Tallahassee FL 32301 – USA, Via G. Dezza n. 24 - 20144 Milan – Italy; 2002.

[7] Mc Laurin, C. A. Hip disarticulation prosthesis. Report 15, Prosthetic Services Centre, Department of Veterans Affairs, Toronto, 1954.

[8] Richards, S., and Richards, N. Artificial Hip Joints: Applying Weapons Expertise to Medical Technology, E&TR 1994.

[9] Klein, M. Using Data in Making Orthopedic Imaging Diagnoses. Advances in Experimental Medicine and Biology 44, 104-111, 2005.

[10] Yusoff, S. F. Knee Joint Replacement Automation Templates. M.sc Thesis, UniversitiKebangsaan Malaysia, Bangi, Malaysia 2009.

[11] Arora, J., Sharma, S., and Blyth, M. The Role of Preoperative Templating in Primary Total knee Replacement. Knee Surgery, Sports Traumatology, Arthroscopy, 13,.3,.187-189, 2005.

[12] Kosashvili, Y., Shasha, N., Olschewski, E., Safir, O., White, L., Gross A., and Backstein, D. Digital Versus Conventional Templating Techniques in Preoperative Planning for Total Hip Arthroplasty. Can J Surg. 52,.1,. 6–11,.2005.

[13] Shapi, A., Sulaiman, R., Hasan, M. K., and Kassim, A.Y. M. An Automated Size Recognition Teqnique For Acetabular Implant In Total Hip Replacement. International Journal of Computer Science & Information Technology (IJCSIT) 3,2,2011.

[14] Hiroyuki, M., Takayoshi, H., Yoshitake, U., Shuji, U., Nobuyoshi, T., and Osamu, N. Evaluation of RF Heating on Hip Joint Implant in Phantom during MRI Examinations. Japanese Journal of Radiological Technology, 66,.7,.25-733,. 2010.

[15] Zhang, W., Titze, M., Cappi, B., and Writz, D. C. Improved Mechanical Long-Permrealiability of Hip Resurfacing Prostheses by Using Silicon Nitride. J Mater Sci: Mater Med , 21,.3049-3057,. 2010.

[16] Scifert, C. F., Brown, T. D., and Lipman, J. D. Finite Element Analysis of a Novel Design Approach to Resisting Total Hip Dislocation. Clinical Biomechanics ,14,. 697-703,.1999.

[17] Phillps, A. T. M., Pankaj, P., Howie, C. R., Usmani, A. S., and Simpson, A. H. R. W. 3D Non-linear Analysis of the Acetabular Construct Following Impaction Grafting. Computer Methods in Biomechanics and Biomedical Engineering. 9,3,.125-33,.2006.

[18] Jonathon, R. Campbell, P., and Mathew, P. E. Metal Release from Hip Prostheses: Cobaltand Chromium Toxicity and the Role of the Clinical Laboratory. Clinical Chemistry & Laboratory Medicine. 51,.1,.213-220,. 2013.

[19] Steens, W., Foerster, G. V., and Katzer, A. Severe Cobalt Poisoning with Loss of Sight after Ceramicmetal Pairing in a Hip—a Case Report. Acta Orthopaedica, 77., 5., 830–832. 2006.

[20] Tower, S. S. Arthroprosthetic Cobaltism Associated with Metal on Metal Hip Implants. BMJ, 344:430., 2012.

[21] Alan, M. K., and Swarts, E. Corrosion of a Hip Stem With a Modular Neck Taper Junction. The Journal of Arthroplasty, 24,.7,. 2009.

[22] Brodner, W., Bitzan, P., Meisinger, V., and Kaider, A. Elevated Serum Cobalt with Metal-on-Metal Articulating Surfaces. The Journal of Bone and Surgery, 79-B,.316-21,. 1997.

[23] Sepehri, A. and Farhang, K. An Extension of CEB Elastic-Plastic Contact Model. Proceedings of the STLE/ASME International Joint Tribology Conference, San Diego, California, 22-24, 10., 2007.

[24] Greenwood, J. A., and Williamsom, J. B. P. Contact of Nominally Flat Surfaces. Proceeding of the Royal Society of London. Series A, Mathematical and Physical Sciences, 295., 1442., 1966.

[25] Chang, W. R., Etsion, I., and Bogy, D. B. An Elastic-Plastic Model for the Contact of Rough Surfaces. ASME J. Tribal, 109,2., 257-263., 1987.

[26] Walker, J. M., and Goldsmith, C. H. Morphometric Study of the Fetal Development of the Human Hip Joint: Significance for Congenital Hip Disease. The Yale Journal of Biology and Medicine, 54.,411-437., 1981.