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A novel analytical approach for determining the frictional moments and torques acting on modular femoral components in total hip replacements

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ABSTRACT

A three dimensional analytical approach was developed to determine the frictional moment vector generated by the relative sliding of the head-cup bearing couple of a total hip replacement. The frictional moment projection onto the femoral neck was also determined over the loading cycle. Predicted frictional moments for nine combinations of bearing materials and diameters were in close agreement with existing *in vitro* data. The analytical method was then applied to simplified gait (lubrication conditions of dry and serum), ISO standard gait and physiological level gait loading cycles. ISO standard gait had a total contact force of about two fold of physiological level gait and there was a corresponding increase in the maximum frictional torque on neck from 0.66 × BW% m to 0.88 × BW% m. For the ISO standard gait, the maximum frictional torque occurred at the same instance of maximum frictional moment and the maximum contact force. In contrast, for the physiological level gait, the frictional torque did not occur at the same instance as the peak load. This suggests that the neck frictional torque is a function of other parameters, such as angle between neck axis and frictional moment vector, as well as the magnitude of the contact force and frictional moment. The developed methodology was able to predict the maximum magnitude and change of directions of moments and the variation of torque at the head neck interface. The data will be useful for experimental studies assessing the fretting behaviour of the head neck junction, by providing appropriate loading data.

1. Introduction

Many contemporary designs of total hip replacement (THR) implants include tapered junctions between the modular neck and stem, and modular head and neck (Campbell et al., 2010). Recent studies report an increase in the number of early retrieval surgeries due to pseudotumours and metallosis, which in part have been caused by fretting and corrosion at the taper junction under mechanical loading in the corrosive physiological environment (Campbell et al., 2010; Cook et al., 2013; Cooper, 2014; Cooper et al., 2013). There have been several studies investigating THR complications associated with fretting wear and corrosion in modular junctions (Bergmann et al., 2010; Bishop et al., 2013, 2008). In these studies, the mechanical loads including hip joint contact force and head-cup frictional moment are understood to be important contributing factors to the fretting corrosion failure. There have also been *in vitro* studies on the fretting corrosion at the head–neck junction to investigate the interrelated effects of the mechanical loads and the corrosive environment around the joint (Duisabeau et al., 2004; Swaminathan and Gilbert, 2012).

In most of the previous *in vitro* studies, the mechanical loads were simplified to a uniaxial force, applied either statically or based on a simplified activity cycle (Baxmann et al., 2013; Duisabeau et al., 2004; Meng et al., 2013; Wang et al., 2008; Zhang et al., 2013). However, physiological loads on modular junctions include 3-axis cyclic forces along with 3-axis cyclic frictional moments resulting from sliding at the bearing couple interface. The projection of the frictional moment onto the axis of the head–neck junction (neck axis) produces a torque on the neck which is transferred within the taper junction. This torque can induce torsional micro-motion at the head–neck interface, which can then lead to multi-axial fretting when combined with the hip contact force. The simplified loading

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Nomenclature (list of symbols)

$\overrightarrow{M}_f$	Frictional moment vector in bearing couple
$\overrightarrow{V}_r$	Rotation vector of the head relative to the cup
μ_k	Coefficient of kinetic friction
A	Contact area (Hertz theory)
D	Lever arm (perpendicular distance between the under pressure differential area and the rotation axis)
r	Radial distance from the centre of under pressure contact area (Hertz theory)
p_0	Maximum contact pressure (Hertz theory)
p	Contact pressure of a point at distance r from the centre of contact area (Hertz theory)
P	Bearing contact force
R^*	Equivalent radius (Hertz theory)
E^*	Equivalent Young's modulus (Hertz theory)
R_i, E_i and ν_i	Radius, Young's modulus and Poisson's ratio, respectively. $i=1$ refers to head and $i=2$ refers to cup
R_{zxy}	Euler rotation matrix for zxy sequence
$R_i(\omega)$	Rotation matrix about axis $i=x, y\ or\ z$ by angle of ω
$c\alpha$ and $s\alpha$	$\cos(\alpha)$ and $\sin(\alpha)$, respectively
C_H^n, C_H^i and C_P	Neutral head coordinate system, head coordinate system at instant i and spherical coordinate system with a z-axis in direction of P, respectively
$x_{C_H^i/C_H^n}$ (or $y_{C_H^i/C_H^n}$ or $z_{C_H^i/C_H^n}$)	x (or y or z) axis of the head coordinate system at instant i with respect to the neutral head coordinate system
$R\left(C_H^{i+1}/C_H^i\right)$	Rotation matrix from C_H^i to C_H^{i+1}
$\overrightarrow{L}$	Position vector of the under pressure points
β	Angle between $\overrightarrow{V}_r$ and $\overrightarrow{L}$
R, θ and φ	Radial distance, polar angle, and azimuthal angle, respectively in spherical coordinate system
M_n	Projection of the frictional moment vector on the neck axis
$\overrightarrow{V}_n$	Unit vector of the neck axis

conditions used in the previous studies may significantly influence the outputs of the fretting and corrosion observed in such experiments.

In vivo mechanical loads acting on hip joints have been measured with instrumented prostheses for a range of daily activities (Bergmann et al., 2001; English and Kilvington, 1979; Hodge et al., 1986; Taylor et al., 1997; Taylor et al., 1998). Although these *in vivo* loads are of great interest, they require costly and complex instrumentation and procedures. Also, these measurements are limited and do not directly measure the torque component acting on the neck axis.

Bishop et al. (2008) used a test apparatus to measure the frictional moments generated by various bearing couples under simulated *in vivo* conditions (cyclic uniaxial loading and uniaxial rotation). Large diameter metal-on-metal bearings generated higher frictional moments than metal-on-polyethylene bearings. Bishop et al. (2013) investigated the influence of lubrication on frictional moments in modern bearings utilising the same experimental apparatus. Large diameter ceramic bearings demonstrated improved friction characteristics under the lubricated condition; however, they could potentially generate high frictional moments if lubrication was compromised. Both of these studies were limited by simplified simulation of body loads and joint kinematics, and in-plane measurements of frictional moment. Developing an analytical method for predicting the frictional moment for bearing couples with different size and material combinations is of great importance.

Previous *in vitro* and numerical studies on THRs failures do not account for the complex frictional moment or multi-axial mechanical loads acting in conjunction with the three dimensional rotational motion of the femoral head (Baxmann et al., 2013; Duisabeau et al., 2004; Zhang et al., 2013). To determine the frictional moment for more physiological loading conditions, a novel three-dimensional analytical approach has been developed that utilises joint kinematics, joint contact loads, and THR material and geometric characteristics. This will provide a means to investigate the effect of forces and moments acting on the bearing couple and the head–neck junction, and to better understand the mechanical environment acting at the head–neck junction during physical activities.

2. Analytical approach

An analytical approach was developed to determine the moment due to friction between bearing couple components ($\overrightarrow{M}_f$) at each instant of a given physical activity. As shown in Fig. 1, $\overrightarrow{M}_f$ is induced in the relative direction of rotation of the head to the cup ($\overrightarrow{V}_r$). The magnitude of $\overrightarrow{M}_f$ can be determined by integrating the product of the differential friction force ($\mu_k pdA$; where μ_k is the friction walker and p is the normal contact pressure according to Hertz contact theory) and its perpendicular distance to the rotation vector (lever arm D), over the contact area (A). Thus, M_f is defined as

$$M_f = \int D\mu_k pdA \tag{1}$$

To calculate this integral, the Hertz pressure distribution, direction of rotation, and lever arm of the differential friction force need to be determined, as well as requiring knowledge of the friction factor.

2.1. Hertz contact pressure distribution

According to the Hertz contact pressure distribution theory, which is suitable for determining the contact pressure between two non-conforming elastically deformed surfaces (Johnson, 1987) such as hard-on-hard bearing couples, the normal pressure (p) at a

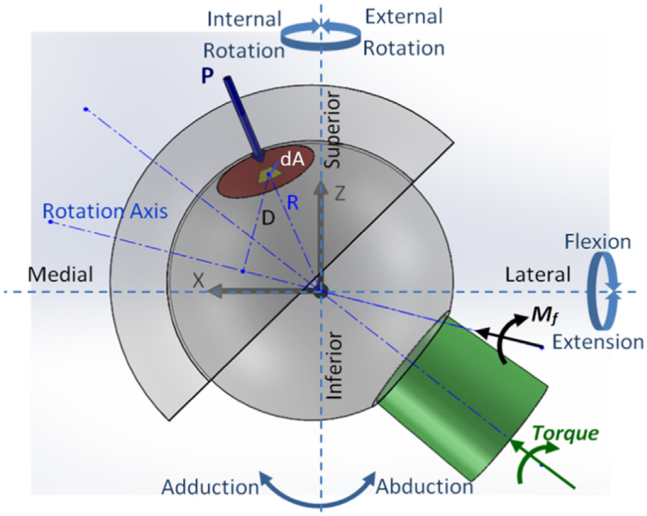

Fig. 1. An antero-posterior view of THR implant with directions of angular motions (left hip, coronal plane).