Tribology International 93 (2016) 239–256

Contents lists available at ScienceDirect

Tribology International

journal homepage: www.elsevier.com/locate/triboint

A review of squeaking in ceramic total hip prostheses

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ARTICLE INFO

Article history:
Received 29 May 2015
Received in revised form
28 August 2015
Accepted 10 September 2015
Available online 25 September 2015

Keywords:
Ceramic total hip arthroplasty
Squeaking
Fluid-film lubrication
Friction
Multibody dynamics methodology
Complex eigenvalue method

ABSTRACT

The occurrence of audible squeaking in some patients with ceramic-on-ceramic (CoC) hip prostheses is a cause for concern. Great effort has been dedicated to understand the mechanics of the hip squeaking to gain a deeper insight into factors contributing to sound emission from CoC hip articulation. Disruption of fluid-film lubrication and friction were reported as the main potential cause, while patient and surgical factors, and design and material of hip implants, were also identified as leading factors. This article summarizes the recent available literature on this subject to provide a platform for future research and development. Moreover, high wear rates and ceramic liner fracture as viable consequences of hip squeaking are discussed.

1. Introduction

The natural hip articulation is one of the greatest natural engineering designs to exist inside the body, Fig. 1. While supporting the entire weight of the body, the hip joint provides a stable and smooth articulation of the lower limb. Although the natural hip joint may provide a lifetime of mobility without any serious problem, chronic pain and disease can affect the hip joint leading to pain and restricted movement. Often affected hip joints are replaced with a biomaterial total hip arthroplasty (THA), Fig. 2. THA restores the physical functioning of the hip joint and reduces pain in most patients, thus improving their social wellbeing and quality of life [1].

The mechanism of THA constitutes a femoral stem fixed in the intramedullary canal of the femur and a ball fixed to the femoral neck of the stem, which articulates in a cup embedded in the acetabular of pelvis, shown in Fig. 3. All components of hip arthroplasty are made of biocompatible materials and biofunctional solutions. The femoral components including the femoral stem and neck are generally made of stainless steel, cobalt-based alloy or titanium-based alloy, while the femoral head is either metal or ceramic. The cup backing can be made of metal or plastic depending on its function. The former used with a plastic cup to secure its fixation to the pelvic bone, whereas the plastic backing is utilized with metal or ceramic cup for absorbing dynamic loads. The most common materials used for bearing surfaces are listed in Table 1 [3]. The mechanical properties and typical roughness values R_a of the above materials are reported in Table 2 [3,4].

THA has revolutionised the treatment of osteoarthritis (degenerative joint diseases), bone tumours, traumas and rheumatoid arthritis. More than 38,000, 80,000 and 200,000 THA procedures are performed annually in Australia, UK and US, respectively with a survival rate of 85–87% after 25 years [5–8]. Due to the difficulty of revision THA and the trauma to patients, it is critical, especially for younger population, that the longevity of the hip implant becomes maximized [9].

Since the early artificial hip joints, around the 1960s, the most used combination is a metal head on a plastic cup (MoP). MoP and ceramic-on-plastic (CoP), also denoted as soft on hard couples, are known to suffer from wear of the plastic part with resultant debris causing osteolysis. In order to reduce the wear rate, alternative hard-on-hard material combinations have been promoted, such as metal-on-metal (MoM) and ceramic-on-ceramic (CoC). However, the presence of potentially cancerous metal ions, developed from wear particles is a serious issue with MoM hip implants [10]. The first CoC bearing was implanted by Pierre Boutin in 1970 [11] in the form of an alumina ceramic ball glued to a metal stem that was cemented in the femur. Alumina ceramic bearings are one of most promising artificial hip joints due to their biocompatibility, high hardness, perfect chemical inertia and low coefficient of friction [12,13]. From the reliability point of view, it has been suggested that surgeons, faced with young and active patients, should consider ceramics as the only safe hard-on-hard bearing surface [14–16].

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http://dx.doi.org/10.1016/j.triboint.2015.09.019
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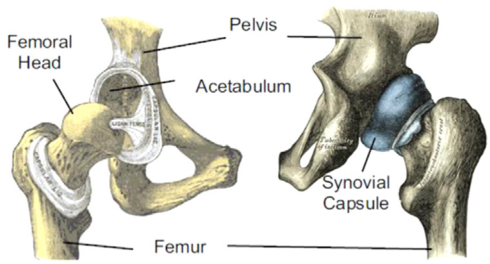

Fig. 1. Anatomy of the hip joint-left: "dissected" joint, right: synovial capsule (adapted from Gray's Anatomy tables) [2].

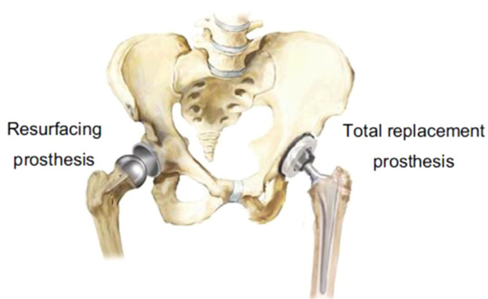

Fig. 2. Total replacement and resurfacing hip prostheses [2].

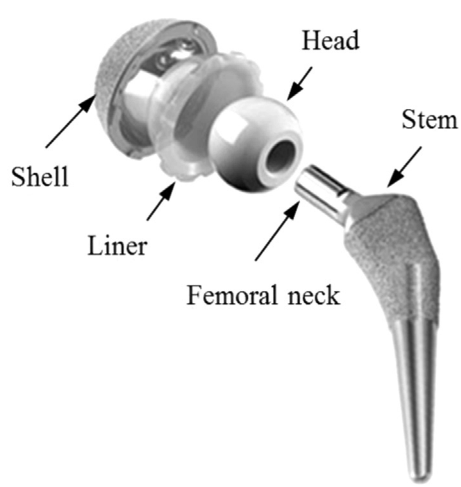

Fig. 3. Main components of an artificial hip joint [2].

Table 1
The most common materials for artificial hip joints [3].

Head	M: stainless steel, CoCr and CoCrMo alloy C: alumina and zirconia
Cup	P: UHMWPEM: CoCr and CoCrMo alloy C: alumina

M: metal; C: ceramic; P: plastic.

Table 2
Mechanical properties of materials and typical roughness values for hip implant components: Young's modulus E, Poisson's ratio υ, average roughness R_a [3,4].

Material		E (GPa)	υ	R_a (μm)
P	UHMWPE	1	0.4	0.1–2.5
M	Stainless steel	210	0.3	0.01–0.05
	CoCrMo	230		
C	Alumina	380	0.3	0.001
	Zirconia	210		

The occurrence of squeaking has however been discussed recently as a cause for concern in THA with CoC bearings [17,18]. From an engineering point of view, squeaking has been associated with friction as the femoral head and cup articulate [19]. However, hip squeaking is multifactorial and important additional factors which may contribute to hip squeaking include: (i) design and materials; (ii) implant position and orientation; (iii) patient factors; and (iv) disruption of fluid-film lubrication and friction. While significant research has been undertaken toward understanding of the mechanism of squeaking, the origins and causes of squeaking still need more investigations. Furthermore, consequences of hip squeaking are unknown and it is not certain if unrevised squeaking hips will result in a clinically adverse outcome to the patient. To date, it has been suggested that high wear rates in artificial hip joints may be associated with hip squeaking [20–22]. It is known that wear of bearing surfaces is a crucial factor in primary failure of all artificial hip joints, influencing their lifetime and performance [23–25]. The intention of this review paper, therefore, is to report the most outstanding work associated with the potential causes and consequences of hip squeaking.

This paper is organized as follows. In Section 2, after defining hip squeaking, a brief description of hip squeaking is provided. The potential factors contributing to hip squeaking are classified into four main groups, namely: (i) design and material; (ii) implant position and orientation; (iii) patient factors; and (iv) disruption of fluid-film lubrication and friction. It has been explained that how these factors affect hip squeaking. Fundamental issues associated with fluid-film lubrication are described, before discussing the disruption of fluid-film lubrication and friction which are main reasons of hip squeaking from an engineering point of view. It is in turn reported that hip squeaking is associated with high wear rates of noisy hips compared to silent hips and ceramic liner fracture. Table 3 summarizes factors linked to hip squeaking by available literature. In Section 3, a comparison of unstable frequencies obtained from clinical data, experimental and computational analyses is presented. Furthermore, the finite element method and multibody methodology as computational approaches for investigation of hip squeaking phenomenon are discussed. Finally, discussing future research directions in this field forms the fourth section of the present article.

2. Hip squeaking

While CoC THA has demonstrated very good clinical performance due to the superior wear resistance and low biological reactivity, the occurrence of audible squeaking in some patients is a cause for concern. Squeaking is defined as an audible sound, 20–20,000 Hz, that occurs during movement of the hip joint, which was firstly described in 1950s [26]. In-vitro, squeaking was also reported by Charnley [27] during the friction analysis of this bearing couple. In fact, hip squeaking has been reported with a wide prevalence rate of 1–24.6% [14,17,18,28–42]. However, it has been reported that no evidence of squeaking observed in their cohort of patients and only eight of patients (6.4%) underwent grinding and clicking noises [37]. In vivo, CoC fundamental squeaking frequencies have been reported in the range of 400–7500 Hz [43]. A spectral view and a fast Fourier transform of squeaking hips analysed in vivo have also been illustrated in Fig. 4. The onset of squeaking was also revealed 14–40 months after total hip arthroplasty surgery [29,38,44–46].

It is worth noting that 15% of squeaking hips stopped emitting noise after a mean follow up of 9.5 years [46]. It has been reported that thirteen squeaker hips out of fourteen stopped squeaking at the last follow up, which its duration was 69.5 months [30], indicating that it could be a temporary pattern. There was no significant difference in patient satisfaction between those with squeaking and silent hips, which showed that squeaking is usually well-tolerated by patients [46]. Furthermore, squeaking is not

Table 3
Studies demonstrating factors associated with hip squeaking.

Authors	Study type	Significant association	No association
Kang [87]	Mechanical study	Negative friction–velocity slope, clearance, material stiffness	
Weiss et al. [115]	Mechanical study	Friction, the level of load magnitude, bearing kinematics, system damping	
Askari et al. [105]	Mechanical study	Stick-slip friction, negative friction–velocity slope, contact force changes	Sprag-slip
Owen et al. [38]	Clinical study		Height, weight, BMI, age, indication
Owen et al. [39]	Meta-analysis	Stryker Accolade femoral stem	
Brockett et al. [83]	Mechanical study	Friction, third body particle, bearing clearance.	
Dacheux et al. [149]	Case report	Ceramic fracture	
Fan et al. [66]	Mechanical study	Femoral stem design , friction	
Hothan et al. [73]	Mechanical study		The head-taper interface
Kiyama et al. [36]	Clinical study	Age, obesity, cup lateralisation, Accolade stem, shortened head length, activity level, pain, satisfaction	Loosening
McDonnell et al. [42]	Clinical study	Range of motion, inclination, anteversion, head size, ligament laxity	Age, height, weight, BMI, gender, satisfaction, stem type
Askari et al. [150]	Mechanical study	Stick-slip friction, negative friction–velocity slope	
Weiss et al. [19]	Mechanical study	Friction-induced flutter instability (whirl), the femoral stem	
Sarialli et al. [85]	Mechanical–clinical study	Edge loading, third body particle, friction, type of motion activity	
Fan and Chen [113]	Mechanical study	A torsional vibration and a flexural vibration of the femoral component	Acetabular component
Sander et al. [84]	Mechanical study	Edge loading, the right combination of load vector and bearing surface conditions, (abduction and contact force)	
Chevillotte et al. [151]	Clinical study	Trident acetabular cup, anteversion, Metal transfer, stripe wear	Age, gender, height, weight
Buttaro et al. [152]	Clinical study		Hip implant design
Chevillotte et al. [55]	Clinical study	Gender, weight, height, activity level,	Age, BMI, neck length, HHS
Haq et al. [35]	Clinical study	BMI, acetabular opening angle, limb length shortening	Age, acetabular anteversion
Kuo et al. [37]	Clinical study	Age, head size, range of motion	Gender, height, weight, BMI, cup size, quality of life, neck length, inclination
Sexton et al. [46]	Clinical study	Height, weight, age, femoral offset, inclination, anteversion, medialisation	Femoral head size, patient satisfaction, BMI, HHS (Harris hip score)
Hothan et al. [57]	Mechanical study	Stem design, assembled stem, axial load	Cup design, bearing clearance
Hothan et al. [69]	Mechanical study		Cup design, assembled cup
Fan et al. [112]	Mechanical study	Friction-induced vibration due to mode-coupling, the femoral stem, neck and head	Ceramic insert
Bernasek et al. [50]	Clinical study	Gender, inclination	
Choi et al. [54]	Clinical study	Head size, gender	Age, height, weight, BMI, cup size, neck length, abduction
Cogan et al. [33]	Clinical study	Association between noise and dissatisfaction	Age, height, weight, BMI, alumina insert thickness
Ki et al. [30]	Clinical study	BMI (body mass index), cup design (Osteonics cup)	Inclination, anteversion,
Parvizi et al. [18]	Clinical study	Neck impingement, Trident acetabular cup, combination of Trident acetabular component and Accolade stem	
Kang [114]	Mechanical study	Negative friction–velocity slope, the femoral stem, bearing stiffness, bearing kinematics, head size, the level of load magnitude	
Currier et al. [110]	Mechanical study	Stick-slip phenomenon, bearing clearance, friction, hip joint velocity	Joint load magnitude, individual components
Sariali et al. [103]	Mechanical study		Micro-separation
Sariali et al. [153]	Mechanical study	Third body particle, high friction, stick-slip	Edge loading, abduction
Weiss et al. [49]	Mechanical study	Self-excited vibrations, high friction, the femoral stem and neck	
Glaser et al. [88]	Mechanical–clinical study	Micro-separation	
Chevillotte et al. [61]	Mechanical study	Material transfer condition, disruption of fluid lubrication	Stripe wear, edge loading, microfracture, joint load magnitude
Mai et al. [31]	Clinical study	Height, neck geometry, V40 neck/Trident combination and C-taper/Trident combination	Age, gender, weight, BMI, indication, head size, acetabular component
Esposito et al. [154]	Clinical study	Type of motion activity	Inclination, patient satisfaction
Restrepo et al. [52]	Clinical study Accolade stem	Age, height, weight, BMI, abduction, anteversion, medicalisation, femoral offset	
Restrepo et al. [64]	Clinical study	Type of motion activity	Pain, functional impairment, ceramic fracture
Swanson et al. [44]	Clinical study	Stryker Trident cup/accolade stem combination, short femoral neck length, rheumatoid arthritis	Age, sex, height, activity level, acetabular component size, femoral head size, BMI, laterality, femoral offset, inclination
Jarrett et al. [56]	Clinical study	Negative quality of life	Inclination, anteversion, leg length, pain, pattern of activity (type of motion activity)
Restrepo et al. [17]	Clinical study	Edge loading, stripe wear, the kinematics of the hip implant	

Table 3 (*continued*)

			Pain, limited function, acetabular component positioning, intervention, abduction, femoral head size, type of femoral stem, impingement, age, height, weight loosening, osteolysis
Keurentjes et al. [29]	Clinical study	Short neck length	Inclination, anteversion, age, gender, BMI, cup size, head size, stem size
Walter et al. [43]	Clinical study	Height, weight, age, edge loading, titanium shells, metallic femoral components, a stiffness mismatch between the shell and liner	Ceramic components, bearing clearance
Brockett et al. [71]	Mechanical study	Bearing clearance, friction	
Murphy et al. [41]	Clinical study		Osteolysis, loosening
Walter et al. [45]	Clinical study	Height, weight, age, anteversion, inclination, impingement, edge loading	
Taylor et al. [89]	Clinical study	stripe wear, edge loading, impingement, inclination	
Jarrett et al. [28]	Clinical study		Pain, anteversion, cup position
Morlock et al. [58]	A case report	Mismatch between the joint bearings, a couple of zirconium oxide and aluminium oxide	

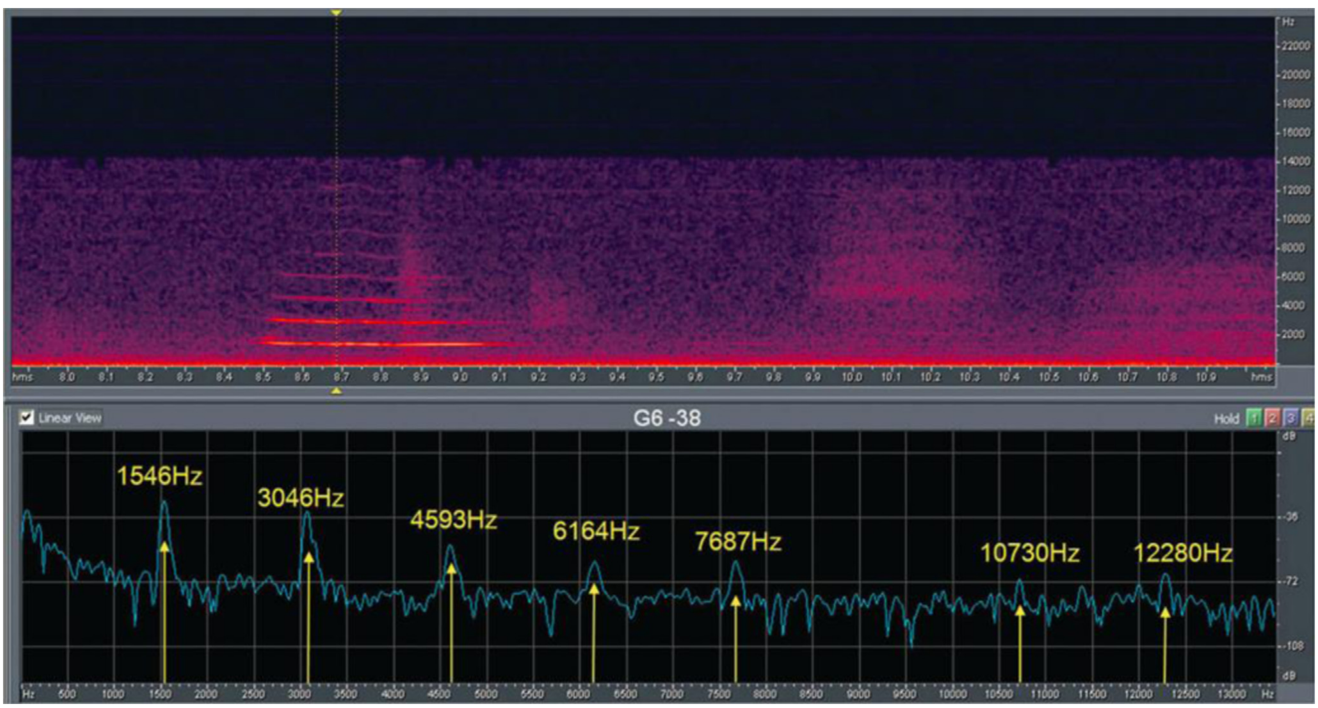

Fig. 4. In the spectral view from the acoustic analysis at the top of the image, the squeak can be seen as a series of parallel lines. A fast Fourier transform of this squeak shows a harmonic series of frequency peaks with a fundamental at 1546 Hz [43].

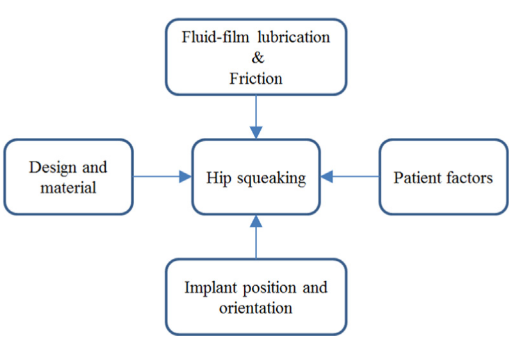

Fig. 5. Diagram showing proposed mechanisms and associations of squeaking in ceramic-on-ceramic total hip replacements.

usually associated with pain, instability and limited hip function [17,41,44,45,47]. Squeaking may be persistent, but more often it is intermittent and tolerable. In some cases, the noise can be avoided by activity modification alone. The incidence rate of revision for squeaking alone is also significantly low, 0.2–0.48% [39,45].

It has been computationally and experimentally demonstrated that squeaking can be associated with the articulation of the femoral head and cup as well as high friction from an engineering point of view [19], namely friction-induced vibration [48,49]. However, hip squeaking is multifactorial as indicated in Fig. 5.

2.1. Implant position and orientation

Implant position and orientation can play a key role in causing squeaking with acetabular component inclination and anteversion, femoral offset and medialization of the acetabular component being the main contributors [45,46]. High or low anteversion and inclination of the acetabular component, which increase the likelihood of impingement and edge-loading, were associated with squeaking, as depicted in Fig. 6 [45,46,50]. Moreover, it has been shown that high prosthetic femoral offset and reduced hip centre medialization of the acetabular component are associated with hip squeaking [36,46]. Increased contact pressure resulting from slight lateralization could also lead to a failure of fluid-film lubrication

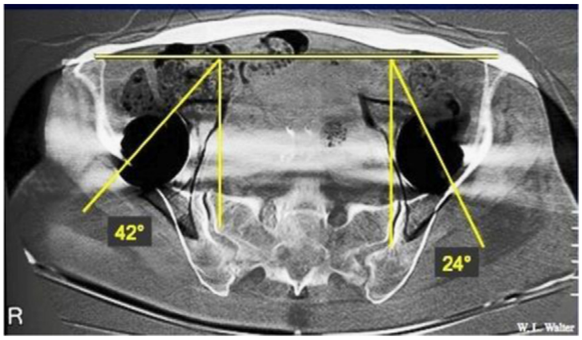

Fig. 6. A computed tomography scan of bilateral ABG II ceramic-on-ceramic hip replacements, the right hip had excessive acetabular anteversion and it squeaked with walking. The left hip with ideal anteversion did not squeak [45].

and increased wear, which could in turn induce squeaking [45,51]. Interestingly, one study revealed that a hip is 29 times more likely to squeak when the acetabular component is positioned out of a defined range [45]. In general, although acetabular component orientation and position were correlated with squeaking, squeaking can also occur when the acetabular component is in the ideal range demonstrating that cup orientation is not the only factor causing hips to squeak [45]. On the other hand, a good number of studies illustrated that there was no direct correlation between the acetabular component position and squeaking [17,52] including anteversion and inclination [17] and femoral offset and stem position [44,52].

2.2. Patient factors

Patient factors, such as age, sex, height and weight, can also influence squeaking [45,46]. Taller, heavier and younger patients with higher levels of activity are significantly more likely to have hips that squeak [36,45,46,53]. Although prevalence of squeaking was also found to be associated with obesity and BMI (body mass index) [30,35,36], Sexton and co-workers reported that they were not associated with audible vibration of THA [46].

Squeaking was also seen more commonly when patients exhibited limb length shortening and rheumatoid arthritis [35,44]. Another patient factor which can be correlated with a higher risk of squeaking is gender. It has been reported that female patients are more prone to squeaking [50], although a number of studies showed it is more frequent in men [54,55]. Interestingly, two cohorts of patients with squeaking CoC hips showed no correlation with age, sex, height, activity level and BMI [29,44]. Furthermore, it was seen that squeaking occurs more in walking, bending, rising and stair-climbing [17,40,45,46,56]. Finally, squeaking is not usually associated with pain, instability and limited hip function [17,41,44,45,47].

2.3. Design and material

Prosthetic design and bearing materials are contributing factors to the prevalence of squeaking [18,44,52,57–59]. Although all types of bearing surfaces exhibit various noises, squeaking has only been described with hard-on-hard bearings [28]. One of the first studies on hip squeaking illustrated that the mismatch of a zirconium head against aluminium cup was associated with hip squeaking and higher surface damage [58]. On the acetabular side, several authors have noted an increased rate of squeaking in the Stryker Trident inserts which has an elevated metal rim, Fig. 7 [18,31,44]. This unique design was proposed to provide protection of the brittle ceramic insert from neck impingement and the material strength increase of the insert [44, 59]. This protective

Fig. 7. A Trident (Stryker Orthopaedics, Mahwah, New Jersey), metal-backed ceramic liner with an elevated rim [21].

rim, however, decreased range of motion by 10–15° [60], leading to metal against metal contact due to neck-rim impingement, Fig. 8, which generates particulate metal debris in the articular surface leading to the disruption of fluid-film lubrication and consequently squeaking [44,61]. Furthermore, the neck-rim impingement also increased the chance of lever out, edge-loading and stripe wear resulting in further damage onto articulating surfaces and squeaking [17,60–63].

The Stryker Accolade femoral stem has a unique V-40 neck design which leads to less impingement according to its smaller neck diameter. Moreover, the titanium–molybdenum–zirconium–iron stem is 25–40% more flexible than that of titanium–aluminium–vanadium [44]. Consequently, these characteristics lead to lower bending stiffness and therefore lower fundamental frequency of the femoral stem, making it more capable of amplifying vibrations generated by hip articulation [44, 64]. Patients with the titanium–molybdenum–zirconium–iron-alloy stem were seven times more likely to undergo squeaking than those with the titanium–aluminium–vanadium–alloy stem [44,52]. After vibrational tests of different hip implant designs, it was confirmed that stem design significantly increased the incidence of hip squeaking, amplifying vibration resulting from a stick-slip mechanism or friction-induced vibration [57,65]. Fan and co-authors reported that shorter, heavier or stiffer stems might limit the possibility of squeaking [66]. A few study also reported higher incidence of squeaking with the Trident cup and Stryker Accolade femoral stem combination than the Stryker Trident cup with other types of femoral stem [18,64].

Short neck length results in smaller range of motion and thus neck-rim impingement due to the tapered nature of the femoral stem [29]. It can also result in soft tissue laxity which can lead to stripe wear and micro-separation and therefore be a precursor of the squeaking sound [29,44,67,68]. The acetabular component size and femoral head size did not directly correlate with hip squeaking [44,46,69]. However, larger diameter ceramic bearings showed higher friction factors, which could make them more susceptible to produce noises than smaller diameter bearings [70]. Moreover, although cup design and the bearing clearance did not show any influence on the dynamic behaviour of the system [69], bearing clearance can affect the lubrication and friction in the bearing articulation [71,72] and the design of the cup can influence the risk of metal transfer and subluxation, leading to squeaking [57]. In addition, a recent study reported that squeaking vibration was not influenced by the head-taper interface [73].

One assumption about squeaking is that it is due directly to independent vibration and natural frequencies of either acetabular or femoral components of the THA, which are correlated with implant design and materials. Although the fundamental frequencies of the metal shells alone ranged from 4.3 kHz to 9.2 kHz, eigenfrequencies of the assembled cup shells after inserting ceramic inlay are above 16 kHz [43,69]. Moreover, natural frequencies of the ceramic femoral head are above the audible human range. These theories can consequently not account for the

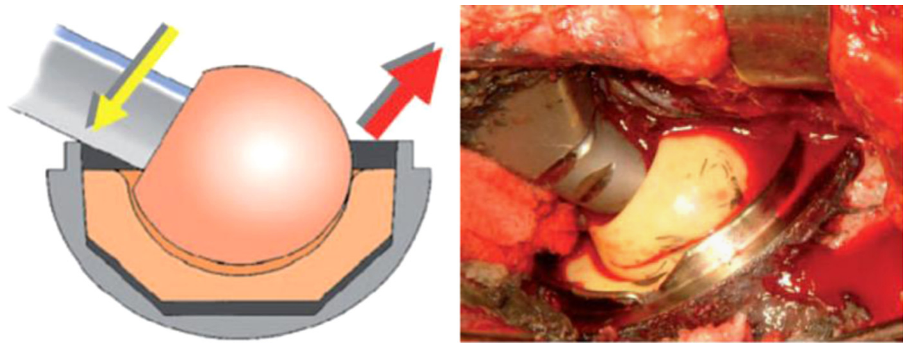

Fig. 8. Impingement caused by elevated rim of Stryker Trident acetabular component. Left plot: femoral neck-rim impingement, which occurs early in the arc of motion potentially resulting in lever out, edge loading, and stripe wear. Right plot: intraoperative view showing notching of the femoral neck and the rim of the acetabular liner and the elevated rim, [44].

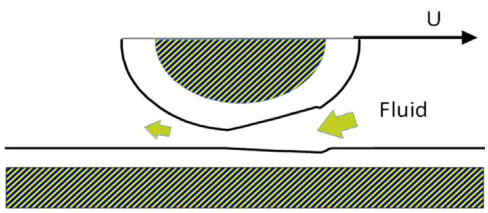

Fig. 9. Schematic of fluid entrainment [76].

observed squeaking [43,69]. However, the vibration frequencies of titanium femoral stem are in the range of 2–20 kHz [43] which are within the audible human range and can contribute to squeaking.

2.4. Fluid-film lubrication and friction

Since the synovial capsule is preserved in THA, the hip implant is lubricated. However, the fluid in this case is more similar to that obtained from diseased joints, which differs from healthy synovial fluid. The characteristics of these fluids are close to the normal bovine serum, which is diluted at 25% indicated in the ISO, which is often used as lubricant for hip implant tests. While these fluids are non-Newtonian and demonstrate piezo-viscosity [74], generally a simple incompressible, Newtonian, isoviscous lubricant model can be assumed. As previously mentioned, squeaking noises are associated with the articulation of the femoral head and cup from an engineering point of view. In what follows, the mechanics of fluid-film lubrication in artificial hip joints is firstly described. In turn, the disruption of fluid-film lubrication and friction are introduced as main causes for hip squeaking.

2.4.1. Fluid-film lubrication

Fluid film lubrication occurs when there is a continuous fluid film separating articulating components. The fluid film thickness must be wider than the average surface roughness to avoid surface asperity interaction and the associated high friction and wear. Fluid film lubrication can be theoretically described by Reynolds' equation, including both the entraining and squeeze film actions. Fluid entrainment occurs when the relative motion between bearing surfaces drags the fluid into the space constituted between them which may separate the articulating surfaces, as observed in Fig. 9.

The relationship among variables affecting minimum film thickness has been demonstrated in the Hamrock and Dowson formula [75]

$$h_{min} = 2.789R\left\{\frac{\eta u}{E'}\right\}^{0.65}\left\{\frac{w}{E'R^2}\right\}^{-0.21} \tag{1}$$

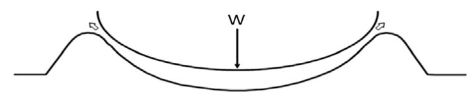

Fig. 10. Schematic of squeeze film formation for a cylinder on a soft flat layer [76].

in which R is equivalent radius of bearing, Eq. (3), η the viscosity of lubricant, u is sliding velocity, w load and E' the equivalent material stiffness, Eq. (4). The equivalent radius of the bearing is defined as the product of the radius of the two surfaces in contact divided by their difference. Consequently, the less clearance/more refined manufacturing tolerances, the more the equivalent radius of the bearing.

Squeeze film lubrication effect occurs when separated surfaces move towards each other very quickly, as illustrated in Fig. 10. The pools of lubricant may be trapped by the contact surfaces, which leak out slowly. The relation of variables affecting minimum squeeze film is materialized in following equation [77]:

$$h_{min} = 2.86R\left\{\frac{w}{E'R^2}\right\}^{0.167}\left\{\frac{\mathrm{d}tE'}{\eta}\right\}^{-0.5} \tag{2}$$

where

$$\frac{1}{R} = \frac{1}{R_1} - \frac{1}{R_2} \tag{3}$$

$$\frac{1}{E'} = \frac{1}{2}\left(\frac{1-\nu_1^2}{E_1} + \frac{1-\nu_2^2}{E_2}\right) \tag{4}$$

in which R_1 and R_2 are the head and the cup radii, respectively, and E_1, υ_1 and E_2, υ_2 denote Young's modulus and Poisson ratio of the head and the cup material, respectively, and t is time variable. Squeeze film formation could occur during walking when heal strikes the ground (heel-strike), Fig. 11, due to ground forces suddenly appeared [78]. During walking, it is feasible that articulating bearings do not come into contact. If, however, the properties of the lubrication break down and the viscosity decreases such as in arthritis, surface contact cannot be longer avoided.

Since the fluid film thickness can be very similar to the average roughness of surfaces articulating, mixed lubrication and boundary lubrication could also take place, even in simple daily activities. In these cases, the bearing surfaces may contact. The lambda ratio (λ) is defined to distinguish the type of lubrication regime, as follows:

$$\lambda = \frac{h_{min}}{R_a} \tag{5}$$

where h_{min} corresponds to the minimum film thickness and R_a,

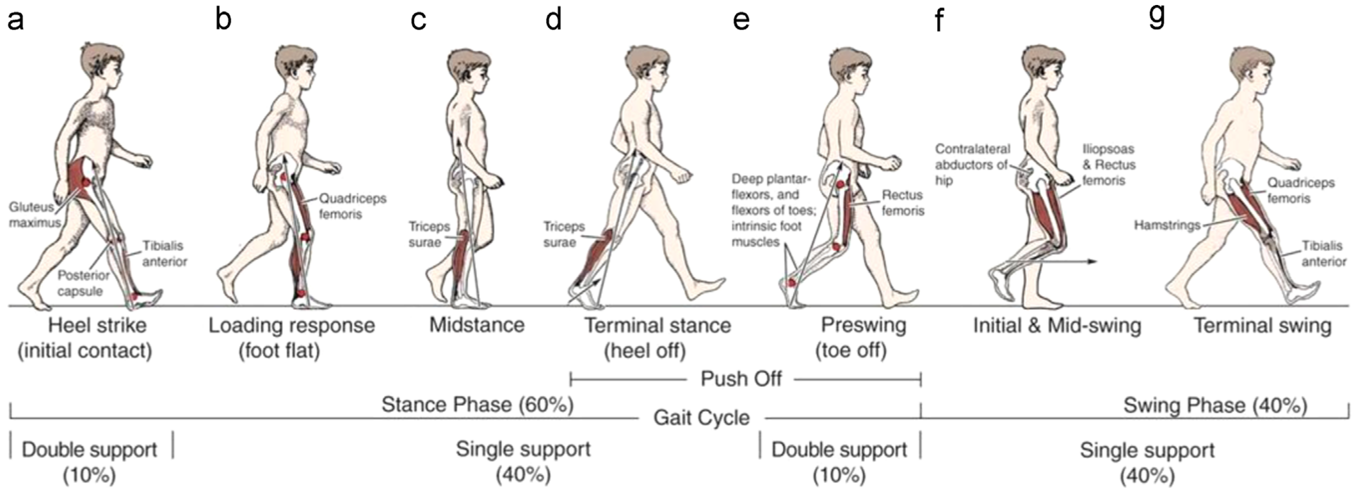

Fig. 11. The gait cycle with the heel strike as the first phase [79].

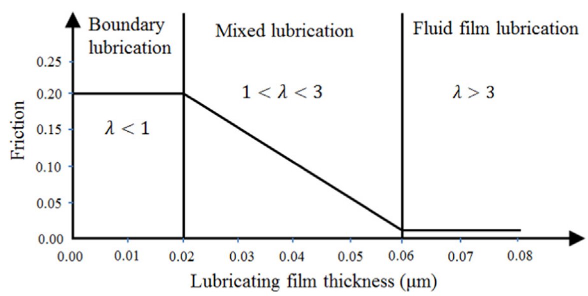

Fig. 12. General relationship between the friction and fluid film thickness [76].

composite roughness of the couple, is

$$R_a = \sqrt{R_c^2 + R_h^2} \tag{6}$$

Here R_c and R_h are roughness of the cup and femoral head, respectively. Once the lambda ratio is evaluated, lubrication regime can be identified as illustrated in Fig. 12. Broadly, it can be stated that MoM THA with a roughness of 0.02 μm and clearance of 0.04 mm is mixed lubricated and CoC THA with a roughness of 0.004 μm and clearance of 0.04 mm is fluid film lubricated [74,80].

2.4.2. Disrupted fluid-film lubrication

Many researchers have suggested that CoC hip squeaking occurs as a result of disruption of fluid film lubrication between bearing surfaces [52,61,81–83]. The fluid film is disrupted by increased surface roughness, particulate metal debris between articulating surfaces, an alteration in the property of synovial fluid and/or abnormal behaviours in prosthetic hip joints such as edge loading and micro-separation. The fluid film is penetrated by large asperities due to high surface roughness and third body particles [61,81,84,85]. The fluid-film lubrication regime is converted to either the mixed lubrication or boundary lubrication according to the lambda ratio described in Eq. (5). Moreover, it was reported that increasing the bearing clearance results in reduced fluid film thickness which can also be concluded by Eq. (5), leading to poor lubrication and consequently squeaking [71,83,86,87]. If the property of synovial fluid changes affecting lubricant viscosity, fluid film thickness decreases, which alters fluid lubrication regime and leads to poor fluid lubrication, Eq. (1).

The non-consistent motion, microseparation, rim-neck impingement and edge-loading of THA articulation, prevents the bearing from producing optimum fluid-film lubrication. Nonconformity of bearing surfaces and inadequate fluid film pressure to bear the femoral head loads can result in stripe wear and third body debris [61,84,85,88,89]. Fluoroscopic studies have elucidated micro-separation between the ball and cup during daily hip motions [90–93]. Moreover, implants with elevated rims, implant mis-alignment and small femoral head size are associated with rim-neck impingement [94,95]. There are also cases of extreme dislocation where the ball exits the socket entirely. Such a dislocation can occur due to neck impingement in which the head is levered out of the cup causing the head to rest upon the socket's rim [96,97]. A relatively vertical cup orientation may also cause edge loading [8]. Edge loading was identified as the first step in a chain of events that leads to CoC squeaking and wear [84].

The fluid film lubrication thicknesses described in Eqs. (1) and (2) are obtained for correctly positioned THA. However, these simulations are not appropriate for adverse loading cases such as edge loading. In this case, there are poor lubrication conditions and extreme contact stresses due to the low conformity of the bearing surfaces [98–100]. Maximum contact pressure caused by edge-loading for a CoC hip implant with 36 mm diameter has been reported as high as 1950 MPa compared to a concentric contact where the pressure was 45 MPa [98]. Moreover, contact pressures for head-liner contact of edge-loaded hard bearings were more than 1 GPa [100]. Edge loading can also cause wear on the surfaces and roughening of both bearing surfaces [84].

While several in vitro studies successfully reproduced squeaking under dry condition [19,57,61,84], it is worth noting that squeaking noises were stopped with adding a small amount of lubricant to a not-lubricated artificial hip articulation [61]. In lubricated conditions, squeaking was replicated by interposing particulate metal debris between the head and liner. Sanders et al. [84] reported that squeaking in lubricated hip prostheses can occur if the right combination of load vector and bearing surface conditions exist such as applying high contact force near the head's wear patch.

2.5. Friction

As can be observed from Fig. 12, friction coefficient increases as the lambda ratio decreases. This is the result of increased surface roughness, particulate metal debris between the articulating surfaces, lower synovial fluid viscosity, increased bearing clearance and abnormal motion behaviour of hip implant components. Generally, coefficient of friction in CoC hip devices is in the range of 0.04–0.13 [80,101–103]. Nassutt and colleagues [101] reported a coefficient of friction varying from 0.104 for no resting duration up to 0.131 for resting duration of 60 s. This increase in coefficient of friction after rest is due to loss of fluid film lubrication and subsequent surface to surface contact when the relative velocity between the bearing surfaces is reduced. This effect tends to make a CoC hip bearing couple act more like non-lubricated bearings [101,102].

It is known that when two surfaces slide against each other, friction develops and acts as a resistance to relative motion. Sliding is an unsteady phenomenon made up of continuous or transient contact resulting in intermittent or cyclical squeaking due to a slight variation in the normal contact load for instance [104–106]. Moreover, frictional force acts like a cross-coupling force linking normal and parallel motions at the contact surface [107]. It is well known that friction can induce vibration in structures owing to instability in the structural system such as the instability due to a surface property for which friction decreases as relative velocity between sliding surfaces increases [48,104,105]. Moreover, there are other sources of instability in structure systems, namely mode-coupling, Sprag-slip, frictional follower forces, stick-slip and material nonlinearity that have all been suggested as possible causes of self-excited friction induced vibration [48,104,105,107–109]. Mode coupling instability is characteristic that frequencies of two structural modes of a system come closer together until they merge and result in a pair of an unstable and a stable mode. The stick-slip exists in most of friction-induced vibration problems as static friction is greater than dynamic friction. Sprag-slip instability is usually characterized by jumping or violation of the system parameters such as contact force and penetration depth during the gait cycle. As friction force follows the displacement during sliding of contacting bodies, it acts as follower force which is a type of instability in the system. Follower forces are well known sources of asymmetry in stiffness matrices and are considered to be responsible for flutter instabilities in a wide variety of mechanical systems. Another type of instability in a system is due to nonlinear stiffness of articulating components, the femoral head and the liner, leading to nonlinear normal contact force.

A possible cause of squeaking in MoM and CoC bearings without lubrication is the stick-slip phenomenon between the head and cup surfaces [110,111]. It has been computationally and experimentally shown that friction-induced vibration is the main reason of hip squeaking [48,49]. In order to consider this issue numerically, a complex eigenvalue method was employed to identify the stability properties of hip implants under laboratory conditions and in a pseudo-in-vivo configuration. However, considerable differences between theoretical and in-vivo results were observed, which could be associated with the choice of boundary conditions [112,113]. These investigations also reported that hip prostheses become unstable when the friction coefficient between components reaches critical values. It was observed that increasing the critical friction coefficient could decrease the occurrence of ceramic bearing squeaking [112,113].

In a theoretical model of a sphere attached to a rotating flexible beam, as a simple model of THA, the bending modes of the beam produced by the dynamic instability under the negative friction–velocity slope was identified as the cause of squeaking [87,114]. An experimental study found that a friction induced whirl vibration led to oscillation behaviour on top of the gross head movement against the liner [19]. This was a micrometre scale elliptical motion inside the liner and the vibrational pattern of hip implants was two-dimensional. However, the hip squeaking frequencies they reported were higher than those found in-vivo. A computational investigation on nonlinear vibration and dynamic behaviour of a CoC hip implant demonstrated that the femoral head had a micrometer/nanometre motion inside the liner [105], illustrated in Fig. 13. This study took the physiological three-dimensional rotation angles and forces into account and found that the vibration of an artificial hip joint had a three-dimensional characteristic. The reported cause of hip squeaking was friction-induced vibration owing to different phenomena such as stick-slip friction, negative-sloping friction and contact force changes. Moreover, friction-induced vibration can increase the sliding distance of the contact

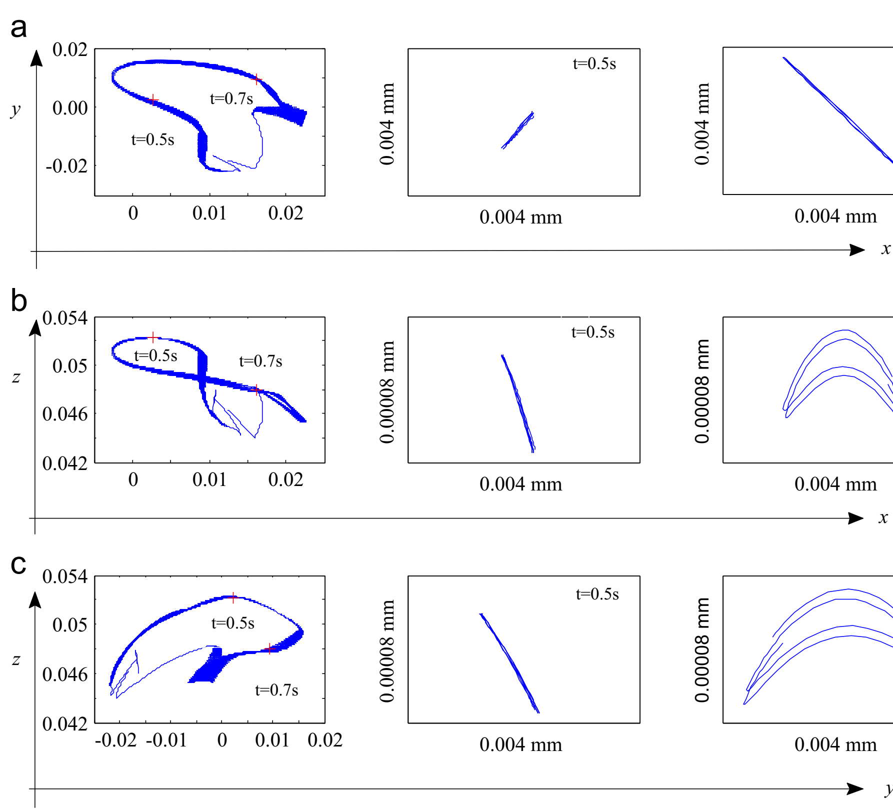

Fig. 13. Contact point track and the vibration of the femoral head in x, y and z directions [105].

point between the head and cup by altering its macro and micro trajectory. It was also observed that friction-induced vibration can significantly affect the contact pressure and joint moments. It was shown that a low friction coefficient, low loads and high system damping decreased the incidence of hip squeaking [115]. Table 3 summarizes the main potential factors related to hip squeaking in available published literature.

2.6. High wear rates of noisy hips observed in vivo

Wear can influence the performance and life expectancy of an implant and has been found to be a key factor in primary failure of artificial hip joints [116,117]. The consequence of wear is often revision surgery to replace the THA with a new one. This is obviously an undesired outcome because of the hardship it imposes on the patient and health budget. Experimental hip simulator and computational studies on CoC and MoM bearings have consistently shown very low wear rates under standard hip simulator conditions which correlates to well-positioned prostheses [23,68,118–125]. However, this has not been confirmed by long-term retrieval analyses [21,126–130]. The standard conditions are defined with the femoral head sitting concentrically with the acetabular cup and the acetabular cup with a clinically equivalent inclination angle of less than 55°. Under these ideal conditions very low wear rates have been obtained. In sharp contrast, CoC and MoM retrievals with elevated wear rates have been associated with steep cup-inclination angle resulting in edge-loading [131–139]. Increased cup inclination angle have been associated with a stripe wear area on the femoral head and an elevated wear rate of alumina CoC retrievals [140]. However, these steep cup-inclination angles exhibited in vitro studies do not lead to high wear levels when tested in-vivo and even the corresponding wear mechanisms [133,135,136,140,141].

Introduction of micro-separation to the gait cycle has however demonstrated edge loading, wear rates and wear mechanisms similar to worn retrievals [123,132,133,139,142–144]. Moreover, the loading and motion inputs affect hip implant wear. Fialho et al. [145] showed that the wear rates occurring during a simulated jogging cycle had a twofold increase compared to those of the walking cycle, due to a significant increase in loading. Considering the effect of different motion inputs on wear prediction of hip prostheses indicated that evaluated volumetric wear under the ProSim simulator and the ISO motion and loading conditions are less than that predicted for in-vivo walking motion [146]. In addition, one study obtained that using a 3D sliding distance increased volumetric wear by 18% compared to a simplified two dimensional flexion–extension analysis [145].

Friction also affects sliding distance and contact stress in artificial hip joints [23,24,105]. It has been reported that the femoral head vibrates inside the cup with micron amplitude within the corresponding collision plane and with nanometre amplitude normal to the collision plane due to friction-induced vibration [19,105]. This can result in a change in the contact point trajectory and contact stress on both a micro and macro-scale, which can

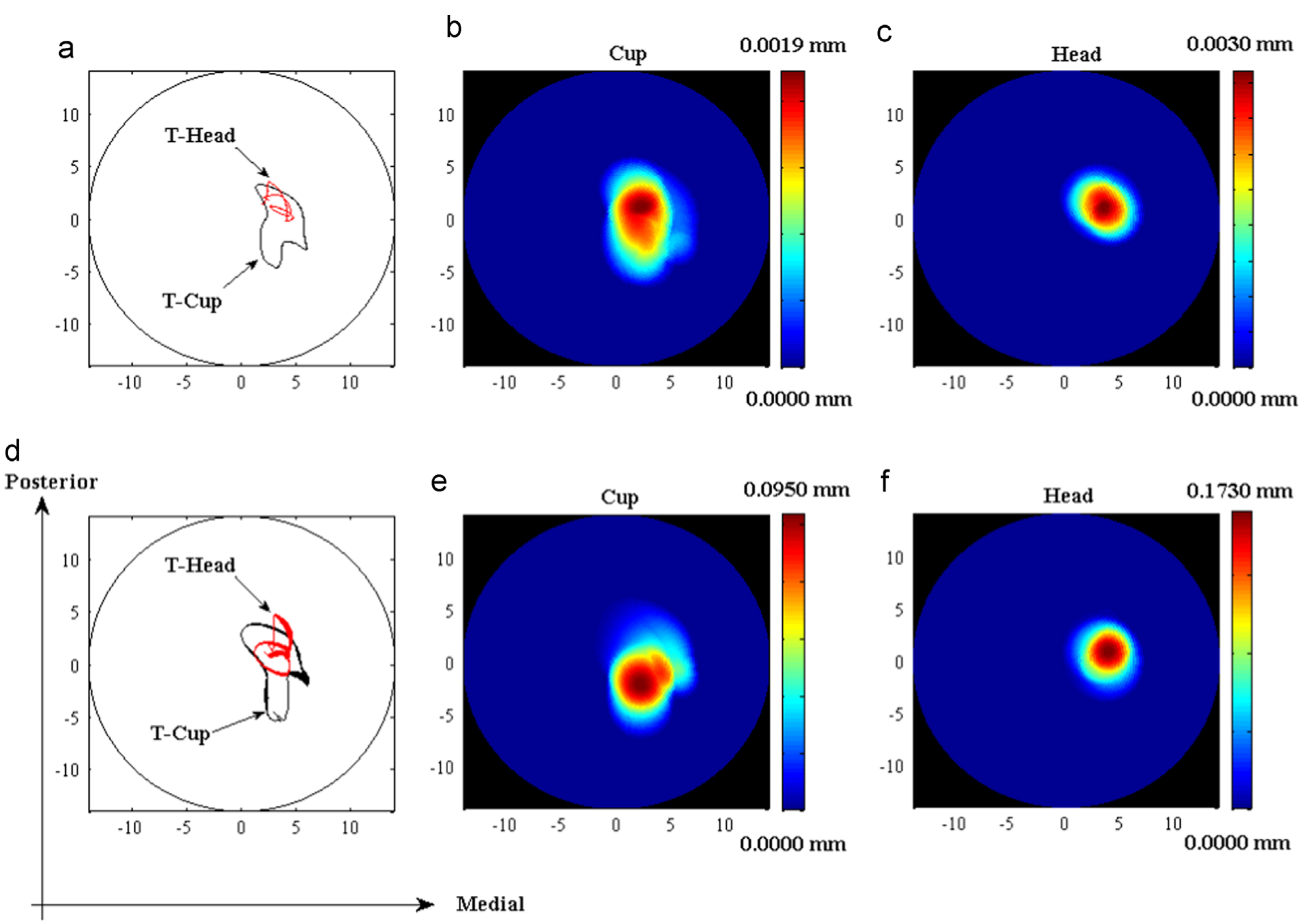

Fig. 14. Ceramic-on-ceramic (CoC) hip implant with three-dimensional physiological loading and motion of the human body with very low friction where volumetric wear is equal to 0.14 mm^3 (top row) and high friction where volumetric wear is equal to 6.9 mm^3 (bottom row): (a) and (d) contact point trajectory on the head and cup, illustrated as T-Head and T-Cup, respectively; (b) and (e) linear wear depth on the cup; (c) and (f) linear wear depth on the head [20].

affect the final wear profile, Fig. 14. Recently, computational simulations showed that friction-induced vibration can considerably increase wear rates in hard-on-hard hip implants. The occurrence of friction-induced vibration in artificial hip joints can therefore be a cause for high wear rates observed in vivo, Fig. 14. High wear rates and cyclic motions of the contact point between the cup and femoral head due to both daily locomotion and friction-induced vibration can also lead to fracture in the ceramic liner of total hip arthoplasty. Recently, a case study reported the occurrence of ceramic liner fracture for two patients with hip pain and squeaking [147].

As discussed previously, the main potential cause for hip squeaking is friction-induced vibration. Moreover, this phenomenon can lead to significant increases in wear rates of hip implants. It can be concluded that noisy hips may experience higher wear rates compared to silent hips. In this regard, a retrieval study, [21], reported that noisy CoC hips had a 45-fold increase in their wear compared to silent hips. Moreover, Askari et al. [20,86,105,148] performed a series of computational studies on hip squeaking and wear prediction of noisy hips showing that noisy hips experience higher wear rates compared to silent hips due to friction-induced vibration, corroborating retrieval studies consistently. Hence, it may be suggested that excessive wear rates are associated with hip squeaking.

3. Computational models

In-vivo and retrieval studies have provided data and information on frequencies at which THA squeaking occurs and contributing factors, such as patient, design/material, implant position/orientation [17,43,44,46,64,88]. In addition, experimental and computational studies have been conducted to understand the mechanism of hip squeaking and figure out how modifying the system can remove squeaking [19,61,105,110,112]. For a comparison purpose of fundamental frequencies of squeaking hips, Table 4 lists two first unstable frequencies with the corresponding method.

Notwithstanding the benefit of analysing in-vivo mechanics using a non-invasive technique and experimental investigations, computational studies have also proved adept at quantifying parametric features. Thus, computational studies play an important role in understanding THA. The finite element analysis as a popular approach for hip implant design and analysis, and multibody methodology to describe bodies by kinematic relations due to vibrational and dynamical nature of hip squeaking are two main methodologies used to simulate noise emission of ceramic hip implants. These methods are briefly described in the following sections.

Table 4
Fundamental squeaking frequencies of ceramic hip bearings.

	First frequency [Hz]	Second frequency [Hz]	Method
Weiss et al. [48]	5	35	Numerical study
Fan et al. [112]	2700	3200	Numerical study
Fan et al. [113]	1843–2050	3300	Numerical study
Ouenzerfi et al. [65]	1759		Numerical study
Askari et al. [86,150]	1700	3400–3800	Numerical study
Askari et al. [105]	2600–3000	5000–5500	Numerical study
Walter et al. [43]	1546	3046	Clinical study
Sariali et al. [85]	2240–2460		Clinical study
Currier et al. [110]	1540–2530	3090–5070	Clinical study
Weiss et al. [19]	3400	10000	Experimental study
Sariali et al. [85]	2600	5300	Experimental study
Currier et al. [110]	2400–3617	4800–7235	Experimental study
Ouenzerfi et al. [65]	2775–3308		Experimental study
Weiss et al. [49]	3350		Experimental study

3.1. Finite element method

The stability of the hip motion equations reflects the likelihood of squeaking of ceramic hip prostheses. Two common techniques for evaluating the stability of a system are (i) transient dynamic analysis; and (ii) complex eigenvalue analysis. A divergent transient solution indicates that the system is unstable. However, this methodology is computationally costly and provides no insight into how the system might be modified to remove the instability. On the contrary, the complex eigenvalue method obtains the system eigenvalues and eigenvectors such that it can be revealed which of a system's vibration modes are unstable. This knowledge enables engineers and designers to deal with the unstable system by means of several control methods. Modal frequencies could be moved by either changing components or adding damping to convert the corresponding unstable modes to stable ones. This method can obtain all unstable frequencies in one run for one set of operating conditions, which would be very difficult to achieve with physical experiments. The complex eigenvalue approach was utilized by Weiss et al. [48] to simulate hip squeaking and they proved that this method was feasible. The method was utilized by other researchers to analyse squeaking noise emitted from ceramic hip arthroplasties during previous years [65,112,113]. Here, the methodology of the complex eigenvalue analysis is introduced briefly. The equation of motion for a vibrating system can be written as follows

$$\mathbf{M}\ddot{\mathbf{u}}+\mathbf{C}\dot{\mathbf{u}}+\mathbf{K}\mathbf{u}=\mathbf{0} \tag{7}$$

where $\mathbf{M}$, $\mathbf{C}$ and $\mathbf{K}$ are mass, damping and stiffness matrices, respectively, and $\mathbf{u}$ is the displacement vector. Due to friction within the contact area, using the Coulomb friction law, the stiffness matrix has specific properties:

$$\mathbf{K}=\mathbf{K}_s+\mu\mathbf{K}_f \tag{8}$$

$\mathbf{K}_s$ is the structural stiffness matrix and $\mathbf{K}_f$ is the asymmetrical friction induced stiffness matrix and μ is the friction coefficient. If the friction coefficient decreases with velocity owing to velocity-dependent friction coefficient, this effect will be added to Eq. (7) by both modified damping and stiffness matrices. The complementary solution to the homogenous, second order, matrix differential equation, Eq. (7), is in the form given below

$$\mathbf{u}=\boldsymbol{\phi}e^{st} \tag{9}$$

Substituting this solution into Eq. (7) yields the complex eigenvalue problem, as follows:

$$s^2\mathbf{M}+s\mathbf{C}+\mathbf{K}\boldsymbol{\phi}=\mathbf{0} \tag{10}$$

To solve the complex eigenproblem, both the damping matrix and the asymmetric contribution of $\mathbf{K}$ is ignored, which yields to

$$(\omega^2\mathbf{M}+\mathbf{K}_s)\boldsymbol{\phi}=\mathbf{0} \tag{11}$$

where ω is an eigenfrequency of the system. Eq. (11) is solved to find the projection subspace formed as N obtained eigenvectors in a matrix, $[\boldsymbol{\phi}_1,\ldots,\boldsymbol{\phi}_N]$. Now, the equation of motion is projected onto this subspace as follows:

$$\begin{aligned}\mathbf{M}^*&=[\boldsymbol{\phi}_1,\ldots,\boldsymbol{\phi}_N]^T\mathbf{M}[\boldsymbol{\phi}_1,\ldots,\boldsymbol{\phi}_N]\\ \mathbf{C}^*&=[\boldsymbol{\phi}_1,\ldots,\boldsymbol{\phi}_N]^T\mathbf{C}[\phi_1,\ldots,\boldsymbol{\phi}_N]\\ \mathbf{K}^*&=[\boldsymbol{\phi}_1,\ldots,\boldsymbol{\phi}_N]^T\mathbf{K}[\phi_1,\ldots,\boldsymbol{\phi}_N]\end{aligned} \tag{12}$$

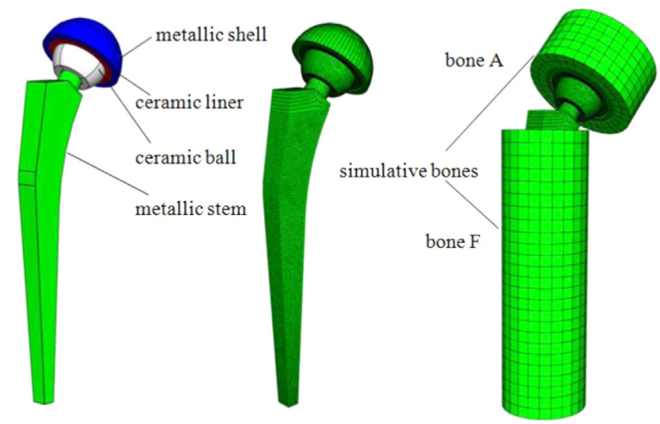

Fig. 15. FE model of a hip endoprosthesis system [112].

Then, the complex eigenproblem is simplified to

$$(s^2\mathbf{M}^* + s\mathbf{C}^* + \mathbf{K}^*)\boldsymbol{\phi}^* = 0 \tag{13}$$

The characteristic equation, Eq. (13), has non-trivial solutions if the matrix in parenthesis is singular. It happens only for complex eigenvalues, $s(s_i)$. The corresponding vectors $\boldsymbol{\phi}_i^*$ are the eigenvectors of the projected system. Finally, the complex eigenvectors of the original system can be acquired by

$$\boldsymbol{\phi} = [\boldsymbol{\phi}_1, \ldots, \boldsymbol{\phi}_N]^T \boldsymbol{\phi}^* \tag{14}$$

Both the obtained eigenvalues and eigenvectors may or may not be complex. Assuming an eigenvalue to be $s_i = \alpha_i + i\omega_i$, where α is the real part and ω is the imaginary part of s for the ith mode, the motion for each mode is written as

$$\mathbf{u}_i = \boldsymbol{\phi}_i e^{(\alpha_i + j\omega_i)t} + \boldsymbol{\phi}_i e^{(\alpha_i - j\omega_i)t} \tag{15}$$

when the real part of the ith eigenvalue, α_i, is positive, the system becomes unstable and squeaking noise emits.

In order to simulate squeaking using commercial FE software, ABAQUS version 6.4 and above provide a complex eigenvalue solution [112]. Moreover, a recent study used the commercial FE software ANSYS Workbench to incorporate the complex eigenvalue method to study hip squeaking [65]. This new capability uses direct contact coupling at the friction sliding interface described by Yuan [155] and there is no need to introduce contact springs at the interface and $\mathbf{K}_f$. The fundamental procedures for applying ABAQUS to perform the complex eigenvalue analysis of ceramic hip prostheses are as follows: (i) Nonlinear static analysis of the ceramic hip endoprosthesis system for applying joint resultant force; (ii) nonlinear static analysis to impose the rotational speed on the femoral components; (iii) normal mode analysis to extract natural frequency without friction coupling; and (iv) complex eigenvalue analysis that incorporates the effect of friction coupling.

Fig. 15 illustrates a CoC THA with a metallic shell, ceramic liner and ball, metallic stem and a simulative bone beside acetabular components, denoted as bone A, and a simulative bone beside femoral components, denoted as bone *F*. Material properties of hip components are listed in Table 5. In previous studies, all materials were assumed homogeneous, isotropic and linear elastic and the element to mesh the model was the 8-noded hexahedral element. The femoral bone and acetabular bone are simulated as cylinders due to the simplification of computational analyses. The hip joint resultant force was set as either 100 N or 1500 N applied at the end of the stem as depicted in Fig. 16 [48,112,113]. A relative rotational motion between the head and the liner was imposed as 1 rad/s around the *Y*-axis to produce friction. Weiss et al. [48] explained that the applied spinning motion is easily extendable to

Table 5
The parameters of prosthetic materials [112]

Materials	Density [kg/m^3]	Young's modulus [N/m^2]	Poisson ratio
Ti6Al4V	4500	1.10×10^{11}	0.3
Ceramic	4370	3.58×10^{11}	0.23
Simulative bone	1932	2.0×10^{10}	0.3

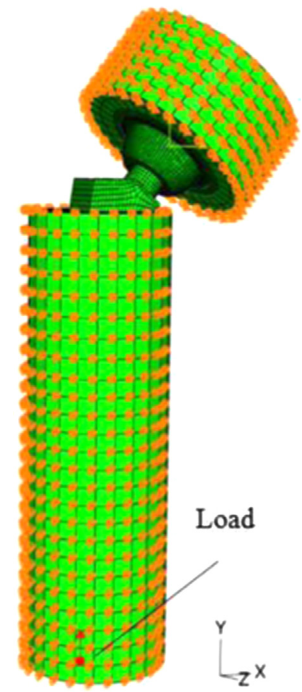

Fig. 16. Loading and boundary condition [112].

the general kinematic motion observed in hip implants in vivo that in fact consists of a superposition of spin and translation. However, there is no study yet to model a hip implant subject to the general kinetic motion, using the finite element method. The bone *A* is completely fixed and the edge of the bone F is supposed to be fixed to assure the original placement of the system. A Coulomb-type friction force with a fixed constant coefficient of friction during each numerical experiment was utilized to allow for a proper identification of unstable parameter-configurations. It should be noted that no FE study to date has taken measured friction–velocity curves into account. Moreover, the effect of slip-stick friction and joint clearance on hip squeaking has not been described using this methodology.

3.2. Multibody methodology

The human body has relatively rigid bones, connected by special joints capable of large anatomical articulations. From a mechanical point-of-view, this description of the human body is similar to that of a multibody mechanical system. However, the human body system is far more complex than the great majority of the multibody systems. Its components have a complex behaviour due to deformations associated with the soft tissues such as the muscles, tendons and ligaments, and due to the complexity of the anatomical articulations relative to the standard mechanical joints [156]. Multibody-based methodologies have been developed in such a way that, besides the representation of mechanical systems made only of rigid components [157], they can also represent deformable bodies [158]. In a broad sense, much of the research

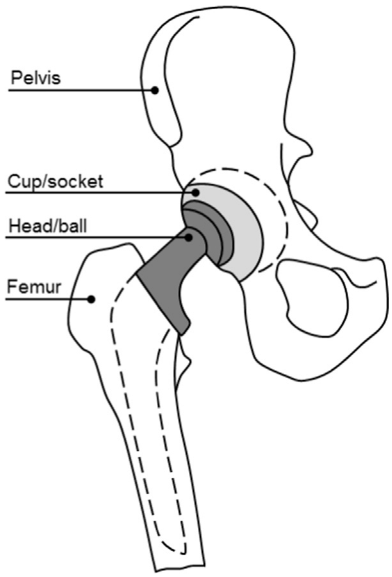

Fig. 17. Schematic representation of the artificial hip replacement.

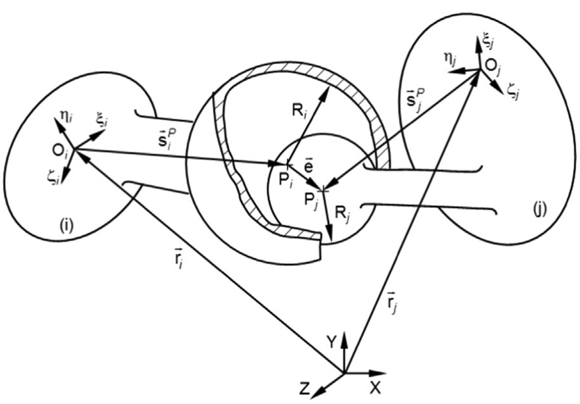

Fig. 18. General configuration of an artificial hip joint in a multibody system [162].

developed with the purpose to simulate daily human tasks is based on the assumption that the joints that constrain the system's components are considered as ideal or perfect joints, such as spherical, revolute and universal joints. Nevertheless, with this approach a decrease in the kinematic and dynamic precision can occur when compared with the living body because the idealized models fail to capture more complex aspects of joint kinematics and dynamics [159].

In the field of multibody system dynamics, computational methods which represent complex phenomena such as contact geometry, friction phenomena, wear and lubrication have been developed [160]. However, the application of these methods in the field of biomechanical system dynamics has been limited. A possible reason is that much of the biomechanical simulation is based upon inverse dynamics, where movement of all degrees-of-freedom is entered into the analysis leading to a presumption of simple joint kinematics. For most applications concerning simple models, this is a reasonable assumption, but for detailed investigations of more complex joints, such as the THA it is not, Fig. 17. In contrast, Askari and his co-authors [150] considered a planar dynamic model of the THA, in which the head and cup are modelled as contacting components [161]. They reported that potential reason for hip squeaking was friction-induced vibration due to stick-slip friction and negative friction–velocity slope. The model developed [148] was extended to a spatial multibody dynamic hip model, taking into account the physiological three-dimensional rotation motions and forces for studying nonlinear dynamics and vibration of THA as well as addressing hip squeaking. Kang [87,114] developed a theoretical dynamic model for the ball joint as a sphere attached to a rotating beam and in contact with a semi-spherical rigid socket. A recent study also investigated friction-induced vibration in THAs by considering contact between the cup and the femoral head [115]. In what follows, the methodology of the multibody methodology is introduced briefly.

It is well known that the equations of motion for a multibody dynamic system with holonomic constraints can be written as [19],

$$\mathbf{M}\ddot{\mathbf{q}} = \mathbf{g} + \mathbf{g}^{(c)} \tag{16}$$

$$\Phi(\mathbf{q}, t) = 0 \tag{17}$$

in which $\mathbf{M}$ is the system mass matrix, $\mathbf{q}$ generalized coordinates of the system, $\ddot{\mathbf{q}}$ the acceleration vector, and $\mathbf{g}$ the generalized force vector containing all external forces and moments. The bodies in the multibody system are interconnected by joints imposing constraints on the bodies' relative motion. Expressing these conditions as algebraic equations in terms of a generalized coordinate and time, t, holonomic kinematic constraints defined in Eq. (17) are introduced. Moreover, $\mathbf{g}^{(c)}$ is the vector of constraint reaction equations, which can be rewritten by means of the Jacobian matrix of the constraint equations ($\mathbf{\Phi_q}$) and the vector of Lagrange multipliers ($\boldsymbol{\lambda}$) as [19]

$$\mathbf{g}^{(c)} = -\Phi_q^T \lambda \tag{18}$$

substituting Eq. (18) in Eq. (16) yields,

$$\mathbf{M}\ddot{\mathbf{q}} + \Phi_{\mathbf{q}}^T \lambda = \mathbf{g} \tag{19}$$

Furthermore, differentiating Eq. (17) twice with respect to time, the constraint equation can be written as follows:

$$\mathbf{\Phi_q}\ddot{\mathbf{q}} = -(\mathbf{\Phi_q}\dot{\mathbf{q}})_{\mathbf{q}}\dot{\mathbf{q}} - 2\mathbf{\Phi}_{\mathbf{q}t}\dot{\mathbf{q}} - \mathbf{\Phi}_{tt} = \boldsymbol{\gamma} \tag{20}$$

where $\boldsymbol{\gamma}$ is a vector function of velocity and position of the system as well as time. As a consequence, both Eqs. (19) and (20) yield a system of differential algebraic equations to be solved for $\ddot{\mathbf{q}}$ and $\boldsymbol{\lambda}$, given by

$$\begin{bmatrix} \mathbf{M} & \mathbf{\Phi_q}^T \\ \mathbf{\Phi_q} & 0 \end{bmatrix} \begin{Bmatrix} \ddot{\mathbf{q}} \\ \boldsymbol{\lambda} \end{Bmatrix} = \begin{Bmatrix} \mathbf{g} \\ \boldsymbol{\gamma} \end{Bmatrix} \tag{21}$$

Eq. (21) can be solved only if the coefficient matrix of Eq. (21) is non-singular. This can be achieved by having a positive definite mass matrix and the Jacobian matrix $\mathbf{\Phi_q}$ full row rank [5]. A general configuration of a hip implant modelled as a multibody system is shown in Fig. 18, and the equations of motion described above addresses its kinetic and kinematic.

Normal contact force and tangential friction force in the articulation surface between the femoral head and cup play important roles in multibody dynamics formulations. There are different approaches to deal with contact-impact events which could be categorised into two main groups, non-smooth dynamics formulation and continuous analysis [163]. In the first group, colliding bodies are assumed to be rigid and unilateral constraints are used to deal with the contact mechanics. On the other hand the continuous methods, also known as either compliant or penalty methods, are considered deformable approaches since the contacting bodies are allowed to deform at the contact zone. Moreover, the corresponding contact forces are evaluated as a function of indentation and compliance of articulating surfaces [163]. As an example, a modified Hertz contact law proposed by Lankarani and Nikravesh [164], which belong to the second group,

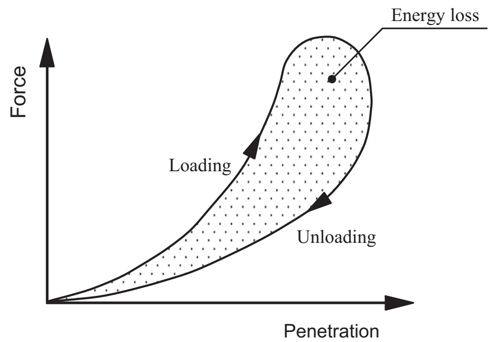

Fig. 19. Force versus penetration [164].

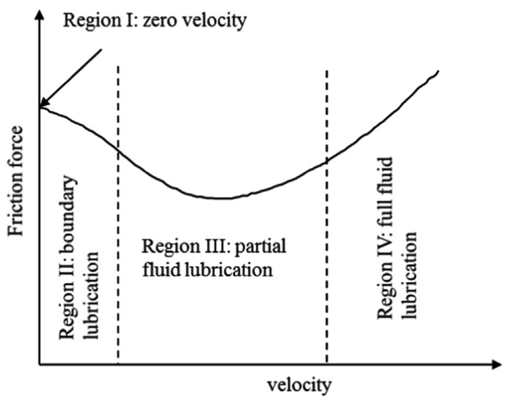

Fig. 20. Stribeck friction–velocity curve showing four regimes [172].

is given by

$$\mathbf{F}^n = -K\delta^{\frac{3}{2}}\left(1+\frac{3(1-c_e^2)}{4}\frac{\dot{\delta}}{\dot{\delta}^{(-)}}\right)\mathbf{n} \tag{22}$$

where $\dot{\delta}$ and $\dot{\delta}^{(-)}$ are the relative penetration velocity and the initial contact velocity, respectively, and c_e is the coefficient of restitution. The generalized stiffness parameter K depends on the geometry and physical properties of the contacting surfaces [10]. This contact model also takes energy loss due to impact and Fig. 19 shows how energy loss during loading and unloading of impact process is captured.

To compute the tangential friction forces, one of the first and simplest friction laws is Coulomb friction law. Coulomb (1736–1806) determined that the frictional force between two bodies which are pressed together with a normal force can be calculated by the product of normal force and friction coefficient. Although the friction coefficient suggested by Coulomb is assumed to be constant with increasing the sliding speed, experimental tests have demonstrated that friction coefficient is a function of the relative velocity, Panovko and Gubanova [165] and Ibrahim [166]. Different friction-speed models have therefore been proposed to take the velocity dependence of friction into account. Moreover, hip implant contact is lubricated and the friction force depends on the friction-speed regimes: (i) boundary lubrication (no dependence on the velocity); (ii) mixed fluid lubrication; and (iii) full fluid-film lubrication, as described in the Section 2.4 and shown in Fig. 12.

Generally, the friction decreases with increased relative velocity until a mixed or full film lubrication is obtained, after which the friction can be constant, increase, or decrease with increasing the sliding speed due to viscous and thermal effects. Stribeck [167–169] suggested a model known as the Stribeck model which can convey the friction behaviour in the different four friction regions. The model can be written as follows:

$$F = (F_c + (F_s - F_c)e^{-(|v|/v_s)^i})sign(v) + k_v v \tag{23}$$

where F is the friction force, v the sliding velocity, F_c the Coulomb sliding friction force, F_s the maximum static friction force, v_s the sliding speed coefficient, k_v the viscous friction coefficient, and i an exponent. This model is represented in Fig. 20. The reason of decreasing friction with increasing velocity in dry sliding metallic bodies was experimentally investigated, and was due to the material softening as a result of high temperatures generated at the contact surfaces [170,171]. The Stribeck model can provide a good representation of the friction between sliding surfaces and it can describe the stick-slip phenomenon and the negative damping effect.

3.2.1. *Operating conditions*

Like the natural joint, THA in-vivo must be able to work under transient and wide range 3D physiological operating conditions, therefore triaxial load and angular velocity should be considered. The most complete database for loading conditions has been provided by Bergman et al., who measured hip forces in nine daily activities, e.g. walking and running, among others. An example of a normal walking gait cycle is depicted in Fig. 21, showing the three components of the hip force and the corresponding hip angles. For simplification purpose, the main (vertical) load component and flexion–extension motion of the walking cycle are usually considered, according to the ISO standard 14242-1, depicted in Fig. 22. The operating conditions are applied to the model as follow: (i) the cup is assumed to be stationary and the forces are applied to the centre of the femoral head; and (ii) the 3D physiological motions are applied to the femoral head.

4. Future research directions

The main limitation of previous computational studies using either finite element method or multibody simulation method is that they have not taken the effect of fluid-film lubrication on the system outcome into account. It can be physically deduced that friction coefficient alters over the gait cycle due to the alteration of fluid film thickness [175]. Moreover, fluid-film lubrication can improve the articulation of the head and cup so it can significantly affect hip squeaking [61]. Therefore, the main future direction should address lubrication effects into numerical formulations to assess hip squeaking.

Using multibody dynamics method, rotational motions and forces of the femoral head are in-vivo inputs so where relevant data is available developed models are applicable. However, there is lack of corresponding information when artificial hip joint is experiencing adverse conditions such as edge-loading and impingement. Hence future work should include whole leg motion with muscles and other soft tissues taken into consideration and then solving equations using inverse dynamics methods to obtain related motions and forces in the case of edge-loading, impingement and micro-separation. Only then can the effect of these adverse conditions on hip squeaking over the gait cycle be investigated.

In addition to these adverse conditions, available multibody models do not simulate elasticity of the contacting bodies, and therefore cannot predict the effect of contact forces and impact on the acetabular components (i.e. distortion of the ceramic insert in the titanium shell) as a potential cause of hip squeaking. Therefore, future work can include the elasticity of contacting surfaces. Moreover, previous studies have only considered normal walking activity while hip squeaking occurs due to other daily activities as

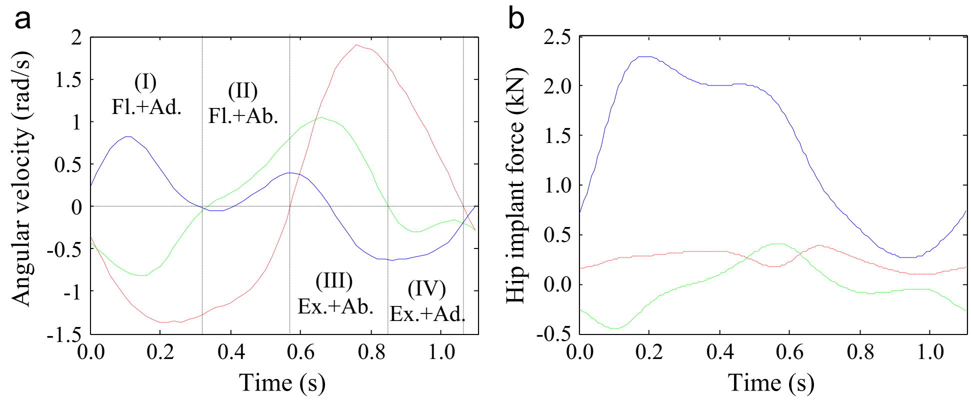

Fig. 21. (a) Angular velocities where ω_z (IER); ω_y (AA); ω_x (FE); (b) physiological adopted forces with f_z (vertical); f_y (A–P); f_x (M–L) for the gait cycle [173].

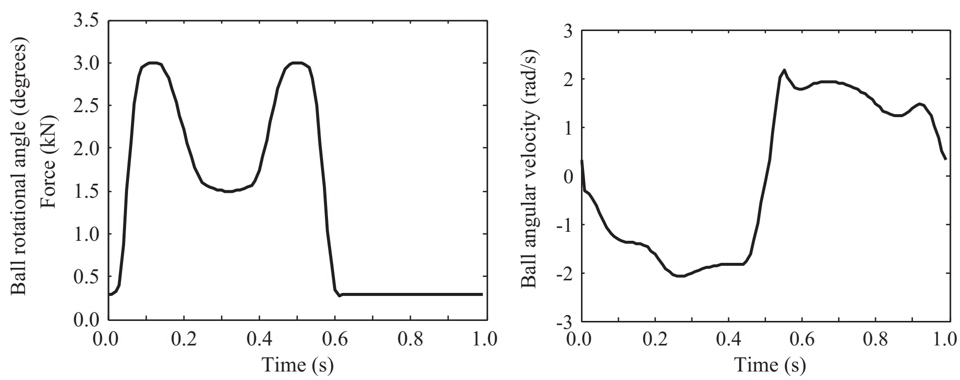

Fig. 22. Left plot shows the vertical load during a normal walking cycle and the corresponding angular velocity is illustrated in right plot based on ISO14242-1 testing standard [174].

well. So the future works could study different daily activities and provide data on hip squeaking due to different daily activities.

Friction was found as the main cause of hip squeaking in ceramic-on-ceramic hip implants. Most of previous studies on simulating hip implant motions employed the Coulomb friction force model which does not account for the Stribeck effect, stick-slip friction and the extremely small displacement of surface asperities. A few investigations have employed a modified friction model including the Stribeck phenomenon to study the dynamics of artificial hip joints, but these models did not account for non-linear pre-slip displacements of the friction interface [87,105,115]. So when relative velocity approaches zero, the friction force is calculated to be zero. In a real sliding scenario, tangential friction force is not zero when very small displacements of surface asperities occur in the contact area. It can significantly affect the kinetic of articulating hip surfaces. An object for future research is to develop a modified friction force model to include stick-slip, Stribeck effect and pre-slip displacement at microscopic level to model nonlinear dynamics and vibration of artificial hip joints, considering hip squeaking.

Although it has been clinically and computationally illustrated that hip squeaking may be associated with higher wear rates compared to silent hips, these results should be validated against controlled physical experiments and gain a better understanding of the mechanism behind articulation and evolution of hip squeaking and wear. Moreover, a recent case report showed ceramic liner fracture for two patients with hip pain and squeaking. There may be an association between squeaking and liner fracture in ceramic hip prostheses and it can be deduced based on an engineering point of view as well. However, this relation has not been proved yet and needs more investigation clinically, experimentally and computationally. Generally, one of main research direction in this field is to discover if hip squeaking has consequences which threaten the life quality and wellbeing of patients and in turn make appropriate clinical and engineering decisions to prevent adverse consequences.

A recent study, which investigated hip squeaking using the complex eigenvalue method, took into account the influence of the soft tissue around the bone on hip squeaking [65]. They illustrated a significant decrease in fundamental squeaking frequencies that are comparable with in-vivo frequencies. It was then discussed that the discrepancy between in-vivo and in-vitro frequencies can be justified by the mass added to the dynamic system due to the presence of the soft tissues. Further experimental and computational studies are required to reveal the effect of both the soft tissues and the bone existing around THA on squeaking. In addition, there are no studies yet to address their damping effects on the vibrational characteristics of the system.

It is now known that the main reason of hip squeaking is friction-induced vibration. As Askari et al. and Fan et al. demonstrated, associated unstable frequencies and vibration modes can be determined using numerical methods [105,112]. From an engineering point of view, ceramic hip implants might be modified by either changing component geometry and/or materials, or by adding damping to convert the unstable modes to stable ones. A recent advance is the development of two new ceramics for THA bearings, namely alumina-toughened zirconia (ATZ) and zirconia-toughened alumina (ZTA). They have shown a 5-fold reduction in the overall wear rate compared to the older alumina ceramics [123,176,177]. Moreover, studies have suggested either addition of damping materials to acetabular components or increase in stiffness of the femoral stem can improve the stability of THA [112,113]. More investigations are required to find feasible and reliable solutions for modifying the mechanical system of total hip arthroplasty to remove squeaking.

5. Concluding remarks

As has been discussed, hip squeaking does not occur in the presence of fluid film lubrication due to non-contact between articulating surfaces. However, the fluid film is disrupted by increased surface roughness, particulate metal debris between articulating surfaces, an alteration in the property of synovial fluid and/or abnormal behaviours in prosthetic hip joints such as edge loading and micro-separation. The fluid-film lubrication regime is converted to either the mixed lubrication or boundary lubrication. Consequently, friction coefficient generally increases which can lead to hip squeaking. Micro-separation, rim-neck impingement and edge-loading are abnormal motion behaviours in THA which prevent the bearing from producing optimum fluid-film lubrication. In these cases, there are poor lubrication conditions and extreme contact stresses due to the low conformity of the bearing surfaces.

In the absence of fluid-film lubrication, bearing surfaces slide against each other and friction develops, acting as a resistance to relative motion. Friction can induce vibration in hip articulation owing to instability in the structural system such as negative-sloping friction, stick-slip, contact force changes, mode-coupling, and material nonlinearity. Moreover, friction-induced vibration can significantly increase wear rates in THA. Hip squeaking may therefore be associated with high wear rates of noisy hips compared to silent hips. High wear rates may lead to the occurrence of ceramic liner fracture, but it needs more investigations to assess as a potential consequence of hip squeaking. There is no more information on hip squeaking consequences.

Acknowledgements

The first author gratefully acknowledges Macquarie University for International Macquarie University Research Excellence Scholarship (iMQRES)-No. 2010017. The second author would like to thank to the Portuguese Foundation for Science and Technology through the project UID/EEA/04436/2013.

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