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REVIEW ARTICLE

Bio-friction

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Abstract: Friction studies in biological systems are reviewed, including synovial joints (cartilage, meniscus), eye, pleurae, fat pad, skin, and oral cavity as well as daily activities associated with shaving, brushing, slip, etc. Both natural systems and medical interventions in terms of diagnoses and artificial replacements are considered. Important relevant biomechanical, physiological, and anatomical factors are reviewed in conjunction with friction studies in terms of both methodologies and friction coefficients. Important underlying tribological mechanisms related to friction are briefly discussed. A unified view on the lubrication mechanism responsible for the low friction in most soft biological tissues is presented.

Keywords: biofriction; soft tissues; friction

1 A brief historical context

The principles of friction have been utilized for centuries in our daily life. For example, journal bearings were used in chariots in China c.2698–2599 B.C. [1]. While in Egypt water or perhaps precious oil was used as a lubricant for transporting an Egyptian colossus from the tomb of Tehuti-Hetep, El-Bersheh, (c.1880 B.C.) as depicted in Fig. 1. This finding was confirmed from a simple estimation of the friction coefficient of 0.23 for the model shown in Fig. 1 and comparison with available modern experimental measurements of about 0.2 between wet wood [2]. Scientific studies on friction began with Leonardo da Vinci, as evidently from a number of his drawings. Subsequently, Amontons, Coulomb, and others made significant contributions to, and laid the foundation for, the current understanding of friction. Nevertheless, as pointed out by the late Professor David Tabor "friction is easiest to measure, but hardest to understand" (Private communication, Dowson).

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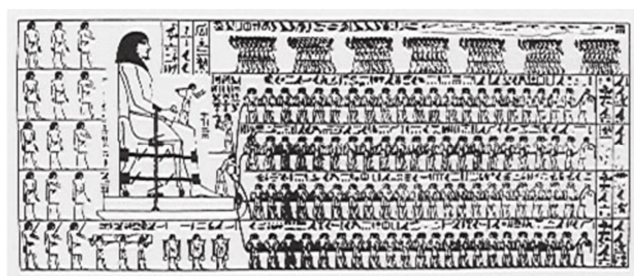

Fig. 1 Transporting an Egyptian colossus from the tomb of Tehuti-Hetep, El-Bersheh, (c.1880 B.C.) [2].

2 Definition

Friction is loosely defined as "the resistance that one surface or object encounters when moving over another" in the Oxford Dictionaries (http://oxforddictionaries.com/definition/english/friction?q=friction). It is interesting to note that the word "friction" was originated in the mid 16th century, "denoting chafing or rubbing of the body or limbs, formerly much used in medical treatment, via French from Latin frictio(n-), from fricare 'to rub'". Bio-friction can be defined as friction applied to biological systems, following on a similar definition of "bio-tribology" by Dowson [3]. It is also noted that "bio-friction*" or "biofriction*" has been used much less frequently in the literature (6 hits searched on the Web of Science on 27th December 2012; as a comparison, "Bio-tribolog*" or "Biotribolog*" was used in 315 hits and "bio-lubricat*" or "biolubricat*" in 180 hits).

Friction forces can generate additional stresses that may become important in contacting bodies. Friction is generally low in biological systems under normal conditions, but can become high under adverse abnormal and diseased conditions. Friction is an integral part of tribology and is closely related to lubrication and wear. In general, friction measurement is much easier to conduct than lubrication and wear. Therefore friction studies are widely carried out to reveal the underlying tribological mechanism. It is important to recognize that a systematic approach should be adopted in friction studies due to the close links between friction, lubrication, and wear. However, it is beyond the scope of the present review paper to address all these tribological aspects and therefore only bio-friction studies are reviewed, with only brief references to related lubrication and wear mechanisms. The importance of friction in normal functions, as well as disease developments in selected natural biological systems, as well as artificial replacements is covered. Nevertheless, for each of the biological systems in consideration, it is equally beyond the scope of the present paper to review comprehensively all the detailed relevant biomechanical and biotribological studies. Other general reviews on lubrication and tribology in biological systems can be found elsewhere [4, 5].

3 Methodology

Friction is not itself a fundamental force but arises from fundamental electromagnetic forces between the charged particles on the contacting surfaces. It is generally very difficult to calculate friction from first principles due to the complexity of these interactions, despite a number of attempts. For example, molecular dynamics simulation has been used recently to predict friction [6]. Instead, friction is usually measured experimentally. Bio-friction studies are usually carried out largely through experimental means due to additional complexities associated with modelling of biological tissues. Such experiments can take simple forms of a simple linear or circular motion where the friction between the two bearing surfaces is measured. In recent times, more and more sophisticated functional simulators have been developed to mimic as closely as possible the physiological environments including loading, motion, and body fluid. Such developments are particularly evident in friction studies of natural synovial joints and artificial replacements as reviewed in Section 4.1.1. Friction is usually quantified as a coefficient of friction (μ). There is a large variation in the reported coefficients of friction in engineering and biological systems due to the complexity of the underlying tribological mechanisms. It is often convenient in engineering to present coefficients of friction with reference to lubrication mechanisms, including fluid-film, boundary or mixed lubrication regime as well as the biphasic lubrication mechanism specifically proposed for biological tissues (Section 4.1.1). Therefore, some values quoted in this paper should be taken as average and representative. For each of the biological systems considered in this paper, a common approach to the literature review was taken; the relevant anatomical structure and physiological/biomechanical environment were briefly mentioned, followed by the discussion on the importance of friction in both normal and abnormal conditions; selected friction studies in terms of both measurement methodologies and representative values of coefficient of friction were presented. Finally the underlying lubrication mechanisms were discussed.

4 Biological systems

Bio-friction studies are reviewed conveniently, according to whether the biological system in consideration is inside or outside the human body.

4.1 Inside the body

4.1.1 Synovial joints

The most important load bearing component inside the human body is the natural synovial joint. Natural synovial joints consist of articular cartilage as the bearing surfaces, bone as the backing materials, and synovial fluid as the lubricant, in a similar way as the journal bearing in engineering as depicted in Fig. 2.

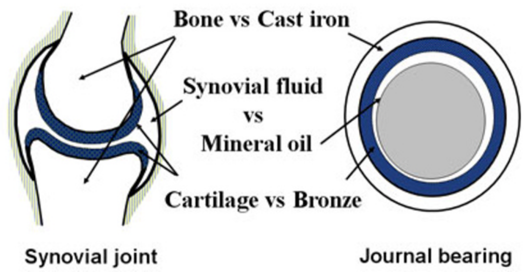

Fig. 2 Comparison of a synovial joint and a journal bearing.

The loading and motion conditions in synovial joints such as the hip are quite complex. Generally, the load during walking is transient and the maximum magnitude can be as high as 4 to 6 times body weight, while the motion is reciprocating with an average angular velocity around 2 rad/s. Most friction studies of synovial joints reported in the literature have utilized small cartilage specimens under simplified loading and motion conditions. There are only a limited number of studies where the whole joint was considered [7, 8]. It is important to measure the friction in synovial joints accurately, since the friction level in these natural bearings is generally low and also because of additional difficulties associated with other soft tissues surrounding the joint and mechanical factors that can contribute to the friction measurement. Unsworth et al [9] developed a pendulum type machine where hydrostatic bearings were adopted to minimize the extraneous mechanical friction. This type of pendulum friction simulator (both free and driven) has been widely used for the friction studies in both natural (hip and knee) and artificial joints (hip) [10].

Healthy synovial joints exhibit friction coefficient as low as about 0.002, despite the fact that they are subjected to a large dynamic load and a reciprocating motion. Table 1 summarizes representative friction coefficients measured in synovial joints under various conditions.

Table 1 Typical representative friction coefficients in synovial joints.

References	Friction coefficient	Comments
11	0.014 to 0.024	Pendulum; cadaveric human ankle joint; boundary lubrication was proposed.
12	0.0053	Arthrotripsometer; dog ankle joint; synovial fluid.
9	0.02	Free pendulum machine with a hydrostatic bearing; cadaveric hip joints; fluid film to boundary was proposed.
13	0.01	Boundary lubrication was proposed.
14	0.01 to 0.5	Cartilage specimen-on-metal; time-dependent friction; biphasic lubrication was proposed.
8	0.02	Driven pendulum machine; bovine knee joint with cartilage-on-cartilage and meniscus; biphasic lubrication was proposed.

The menisci are known to play important roles in the normal function and the development of diseases such as osteoarthritis in the knee joint. Pickard et al. [15] compared the time-dependent friction between bovine meniscus and cartilage, both against a stainless steel plate and found that the friction coefficient for the meniscus tissue was higher, particularly during the early stage of loading. The effect of the meniscus on the friction of bovine knee joint was investigated by McCann et al. [8]. It was shown that the removal of the meniscus significantly increased the friction coefficient between the cartilage surfaces from 0.02 to 0.05 as a result of the increased contact pressure. Baro et al. [16] also found a similar friction coefficient on the order of 0.02 under migratory contacts and further showed that the femoral apposing surface tended to give lower friction than the tibial counterpart. It is generally accepted that a migratory contact allowed the re-hydration of the biphasic materials and recovery of the fluid-load support.

The low friction inside synovial joints is generally accepted. However, the underlying mechanism is still not clear. It is probably a combination of various effective lubrication mechanisms, ranging from boundary, mixed in the form of biphasic lubrication to fluid-film lubrication as discussed below [17]. Under normal conditions, the softness of articular cartilage promotes the formation of fluid films and this reduces friction markedly. Even when a fluid-film lubrication regime is not possible, boundary lubricating constituents of synovial fluid often reduce friction to a level that is not much different from that under a full fluid film lubrication condition [11]. Another friction-reduction mechanism is biphasic lubrication in articular cartilage, which consists of both fluid and solid phases. Immediately after loading, the fluid phase inside cartilage is pressurized and therefore the majority of the load is carried out by the fluid phase, resulting in low friction [14, 18]. As time increases, the load is transferred to the solid phase and friction increases. Under a prolonged period of loading, boundary lubrication may act as an effective mechanism to limit friction in synovial joints. Other lubrication mechanisms proposed for articular cartilage include hydration or brush, which may be related to biphasic lubrication [19] or boundary lubrication [13].

The hydration lubrication mechanism in articular cartilage has received significant attention recently. The essence of the hydration lubrication mechanism is a "surface amorphous layer", also described with different names such as "gel layer", "hydration layer", or "brush layer", in which the condroitin- or keratan sulphates composing the leafs of the proteoglycan subunit are hydrated [20]. Recent studies by Klein and colleagues [21] have revealed the remarkable ability of phosphatidylcholine liposomes to reduce friction coefficients on atomically smooth mica surfaces to exceedingly low values of around 10^{-4} under physiologically relevant pressures.

Consideration of friction between articulating surfaces has played an important role in the development of artificial hip joints. The hip replacement designed by the late Sir John Charnley utilised a material combination with a minimum friction coefficient under boundary lubrication (e.g., Teflon (PTFE)). Combined with a small femoral head diameter of approximately 22 mm, this gave a low frictional torque; the principle for the Low Friction Arthroplasty (LFA). Later on, PTFE was replaced by high density polyethylene and then ultra high molecular weight polyethylene, however, the principle of LFA has remained. Subsequently, it has been shown by Wroblewski et al. [22] that the loosening rate of acetabular cups was reduced for a thicker polyethylene cup, particularly when the linear wear penetration exceeded 1 mm. This has been explained on the basis of the shearing stress at the cup/cement interface resulting from the frictional torque generated at the articulating surfaces. A decrease in the outside diameter and an increase in the linear wear penetration resulted in an increase in the shear stress and likelihood of loosening. Friction may also have played an important role in the clinical performance of large diameter metal-on-metal hip implants. The large frictional torque in these devices under adverse lubrication conditions due to edge loading and micro-lateralisation may be responsible for both the cup and the taper connections loosening and clinical failures identified recently [23, 24]. Typical friction coefficients in artificial hip joints with different bearing surfaces are summarised in Table 2.

4.1.2 Fat pad and tendon

Fat pads are masses of encapsulated adipose tissue, commonly found and strategically located within the human body to provide mechanical advantage to the musculo-skeletal system. Fat pads consist of water, collagens and proteoglycans as the extracellular matrix, and numerous unilocular adipocytes (fat cells) that are swollen with lipid. Fat pads play an important role in reducing friction in the musculo-skeletal system as reviewed by Theobald [27]. Under adverse conditions, high friction may lead to abnormity and consequently pain. For example, one of the common causes of anterior knee pain is known as the patellar tendon lateral femoral condyle friction syndrome. This is caused by patella maltracking resulting in the impingement of the superolateral aspect of Hoffa's fat pad between the inferior patella and the lateral femoral condyle.

The friction between fat and bone from bovine tissue was measured by Theobald et al. [28]. A typical coefficient of friction of 0.01 was reported. These authors also adopted the Sommerfeld analysis commonly used in engineering and found that predominant hydrodynamic lubrication was present in their experiments. They further suggested that one of the functions of fat pads associated with

Table 2 Typical friction coefficients (factors) for various bearings for artificial hip joints in the presence of bovine serum [25, 26].

Bearings	Friction factor
UHMWPE-on-Metal	0.06–0.08
UHMWPE-on-Ceramic	0.04–0.06
PEEK-on-Metal	0.35
PEEK-on-Ceramic	0.36
Metal-on-Metal	0.10–0.18
Ceramic-on-Ceramic	~0.04
Ceramic-on-Metal	~0.04

subtendinous bursae and synovial joints should be to generate a hydrodynamic lubricating layer between the opposing surfaces.

Tendons transfer muscular forces around the joint, facilitating joint motion. Tendons can be subjected to either tension (i.e., mid-substance) or compression (i.e., fibrocartilaginous). High friction in tendon has previously been reported in association with cumulative trauma disorders such as carpal tunnel syndrome and tendonitis as well as tendon suturing failure [29]. The friction between a canine flexor digitorum profundus tendon and its pulley was quantified by Uchiyama et al. [30] using two force transducers connected to each end of the tendon. A frictional coefficient of the canine flexor tendon-pulley was found around 0.016.

Theobald et al. [31] reported experimental data describing the friction characteristics of the tensile and compressive regions of bovine flexor tendon against glass using a pin-on-plate tribometer. Under physiological conditions, the tensile tendon region was found to be capable of generating elastohydrodynamic lubrication, with a coefficient of friction around 0.1 mainly as a result of viscous shearing in a fluid-film lubrication regime. The coefficient of friction in the equivalent region of compressive tendon was measured as 0.008, in the mixed/boundary lubrication regime. The surface-bound lubricin (a glycoprotein present in the synovial fluid that specifically binds to the surface of tendon, articular cartilage, etc.) was also found in the compressive region, which has been shown to be an effective boundary lubricating constituent responsible for minimising the friction in the mixed/boundary lubrication regime. However, such a lubricating mechanism has not been found in a number of synthetic grafts [32].

4.1.3 Pleurae

Friction also plays an important role between the normal function as well as disease developments between the lung and the chest wall. The pleurae consist of a double membrane with a monolayer of mesothelial cells, covering the lung (visceral pleura) and lining the chest wall (parietal pleura) [33]. There is a potential space between the double membrane, the pleural cavity, where a lubricant known as pleural (serous) fluid is found. It is important to ensure effective lubrication between the pleural membranes and low friction and minimum shear stress between the two sliding membrane surfaces during breathing [34]. However, under some adverse conditions, friction may be significantly increased, potentially causing damage to the tissue surfaces as well as producing an audible sound. This latter phenomena has been used to diagnose pleurisy and other conditions affecting the chest cavity such as pneumonia and pulmonary embolism, as commonly known as a pleural friction rub, or simply pleural rub as the pleural layers are inflamed and whenever the patient's chest wall moves during inspiration and expiration.

The measurement of friction in pleural surfaces has largely been carried out *in vitro*. The experimental results have been inconsistent, mainly due to the simplified apparatus and external conditions and the preparation of samples. A simple inclined plane was used in early experiments to measure starting coefficients of friction of lung sliding on the inner chest wall and a typical value at approximately 0.2 was found [35, 36] studied rabbit lung sliding on chest wall pleura with pleural liquid as lubricant in an *in vitro* set-up. They found the starting coefficient of friction increased from 0.086 to 0.122 as the period of stationary contact increased from 5 to 30 s. It is interesting to note that such a time-dependent friction characteristic is consistent with that observed for articular cartilage as discussed in Section 4.1.1. Under dynamic oscillating conditions representative of physiological velocities and normal forces, the average value of the coefficient of kinetic friction was constant at 0.019. Furthermore, the friction characteristics measured in both these experiments were broadly consistent with boundary lubrication, with substantial contact between the surfaces. However, other experimental results were more consistent with a full fluid film lubrication regime [37]. Friction was measured in a rotational tribometer during steady state sliding between mesothelial tissue from the peritoneal mesothelial surface and smooth glass lubricated with normal saline. The friction characteristics were found to be consistent with a progression of lubrication regimes from mixed to fully developed hydrodynamic lubrication. Potential differences between these studies were the apparatus and the samples used, as pointed out by Loring and Butler [38]. This highlighted the importance of large scale conformation differences among tissue samples that promoted load support and reduced friction to a variable extent. Alternations to the natural system, i.e., blotting with filter paper, can significantly increase the friction and damage the pleural surface [39].

As with so many soft biological tissues, there are a number of potential lubrication mechanisms responsible for the low friction in the pleural surfaces. Elastohydrodynamic lubrication at microscopic scales has been proposed to be responsible for effective lubrication and low friction between parallel pleural surfaces. The asperities on the pleural surfaces and subsequent deformation promotes hydrodynamic load support and separates the two sliding surfaces [40]. Boundary lubricating properties of the pleural surfaces are also responsible for reducing friction. A number of boundary lubricating constituents have been identified, including surface active phospholipids [39, 41], again similar to those found in synovial fluid. However, the exact lubrication mechanisms remain speculative and controversial.

Similar to the lungs, the heart and intestines would probably work in a similar manner. They all need to change their shape and size and slide against the chest wall and other organs to function normally. A similar effective lubrication mechanism may be operative during this process to provide little friction and without apparent damage or wear. Destruction and damage of the surfaces may elevate friction and result in diseases in all these soft tissues.

4.1.4 Eye

Normal functions of the eye depend on effective lubrication and minimum friction and wear between the cornea and the eyelid. The cornea is approximately spherical in the central portion, however, its surface is not smooth. The surface topography on the cornea has been found to have microridges up to 0.5 μm high [42]. However these microridges are covered with a mucus gel so the effective roughness may be much less. Tear films also play an important role in the lubrication of the eye. Tear films have three distinct layers: the outermost being a lipid (fatty, oily) layer having a thickness of about 0.1 μm, the middle layer being an aqueous layer of 7–10 μm thick and low viscosity, and the innermost being a viscous mucous layer to adhere to the cornea surface. The major biomechanical function of the eye is blinking, which was studied in detail by Hayashi [43]. Blinking occurs once every 5 s on average. It takes about 0.08 s and 0.17 s during closure and opening respectively. During closure the upper eyelid moves down with an approximate speed of 0.15 m/s. The normal load between the eyelid and the cornea ranges from 0.2 to 0.25 N. Loss of lubrication and increase in friction can result in dry eye syndrome, either because of less production of tears or more watery tears than oily or both. High friction can result in high shear stresses, and inflammation and damage to the anterior tissues, leading to inconvenience to patients and scratching and burning of the eyes. Dry eye syndrome may be treated by using artificial tear drops.

Direct friction measurements in the natural eye have been rather limited and most friction measurements have been done on tear drops and contact lenses. Cobb et al. [44] developed a low load friction measuring apparatus and determined the coefficient of friction between a glass pin and an intact layer of human corneal epithelial cells of the order of 0.05. Furthermore, they showed a direct relationship between the coefficient of friction and the extent of cell damage. Contact lenses are widely used to correct eyesight. The introduction of a contact lens in the eye results in two biotribological interfaces: the post-lens between the posterior surface of the lens and the eye surface (cornea) and the pre-lens between the anterior surface of the lens and the eye-lid, with the latter being more critical in terms of friction. Friction from the pre-lens interface of soft contact lenses has been measured in a number of studies. Rennie et al. [45] used a microtribometer to measure friction in a number of commercially available contact lenses slid with a glass pin under a wide range of contact pressures and speeds. The friction force was found to consist of three components: viscoelastic dissipation, interfacial shear, and viscous shearing. The coefficients of friction were found to vary from 0.025 to 0.075. Another similarly sophisticated friction apparatus was developed by Ngai et al. [46], where a silicone rubber eye-form that retained the contact lens was slid against a smooth reciprocating flat glass plate.

lubrication mechanism in the natural eye, as well as in the presence of a contact lens, has been studied for a long time. Early studies by Ehlers [47] suggested boundary lubrication, however Holly and Holly [42] proposed an alternative hydrodynamic lubrication mechanism due to the relatively thick tear film discussed above. Extensive studies have been carried out to measure the tear film thickness in the eye, and the post-lens and pre-lens tear film thicknesses in the presence of a soft contact lens. At the same time, a number of theoretical lubrication modelling studies on contact lenses have also been carried out [48]. All these experimental and theoretical studies gave some evidence supporting the role of elastohydrodynamic lubrication in contact lens friction, broadly in agreement with the friction studies discussed in this section.

4.1.5 Oral cavity

Human oral cavity is quite complex, consisting of both hard and soft tissues such as palate, chin, teeth, tongue, mucosa and glands as well as the temporomandibular joint (TMJ) which connects the upper temporal bone with the lower jaw bone. The TMJ can be considered as a synovial joint and therefore expected to behave similarly to other synovial joints such as the hip and the knee as reviewed in Section 4.1.1. All the soft tissues in the oral cavity are covered with mucosa, which is lined by stratified squamous epithelium with topographic differences that correlate with masticatory demands [49]. Another important element in the oral cavity is saliva. Understanding of the lubricating properties of saliva may help develop saliva substitutes [50] to treat "dry mouth" symptoms. As an organ, the main function of the oral cavity is closely related to speech and food processing. Therefore, friction can be expected to play an important role in the oral cavity. For example, during chewing, the movement of the teeth with the lubrication of saliva or food slurry results in friction and wear. Various names have been used to describe particular examples of frictional keratosis in the oral cavity from excessive force. Brushing the teeth may cause toothbrush keratosis, the constant rubbing of the tongue against the teeth may lead to tongue thrust keratosis and injuries to the oral mucosa may result in frictional keratosis. Another important aspect in the oral cavity is related to oral processing. The overall behaviour of a food in the mouth depends on how the food interacts within the oral environment. A number of processes are involved when food is prepared for swallowing in the mouth, including the mechanical breakdown of solid pieces into smaller fragments, enzymatic reduction of starches into sugars, molecular interaction with micro-organisms, and mixing with saliva. This requires a wide range of complex movement of the teeth and the tongue and different types of shear, tensile and compressive deformation. Furthermore, there is considerable interest in the possible link between texture, friction, rheology, and human perception of foodstuffs, such as creaminess and astringency [51], in a similar manner as the skin discussed in Section 4.2.1.

A wide range of methods has been applied to measure friction during oral processing of food as well as producing food as reviewed by Goh [49], including the linear friction sledge, the pin- or ball-on-disk tribometer as well as rheometers with specific friction attachments. The important considerations for the contacting surface may include hydrophobic or hydrophilic, structures with pillars to simulate the papillae on the tongue and in some cases using animal tongues. The effect of surface structure on frictional behavior of a tongue/palate tribological system was investigated by Ranc et al. [52] under both dry and oil and aqueous solution in a reciprocating motion sliding tribometer. The friction was shown to be strongly affected by the topographical structure of the contacting surfaces. The effect of brushing on adsorbed salivary conditioning films and friction was investigated by Veeregowda et al. [53] using colloidal probe atomic force microscopy under different modes of brushing (manual, powered, rotary-oscillatory or sonically driven). It was found that different modes affected the friction and the mode of lubrication.

The coefficients of friction of oral tissue, including teeth, have been shown to range from about 0.004 to 0.45, depending upon the external environment and conditions of load, sliding speed, and counterface as summarized by Dowson [54]. Coefficients of friction in the presence of whole mouth saliva range from 0.02 to 0.2, with clear evidence of both boundary and mixed lubrication characteristics. Under certain conditions, when softer substrates were used, a transition from mixed to fluid-film lubrication was possible, with a minimum coefficient of friction of around 0.004 in the Stribeck curve. Harvey et al. [55] performed surface balance experiments on human whole saliva absorbed to molecularly smooth mica substrate and found a coefficient of fiction of 0.24 and 0.46 for the unrinsed and rinsed systems, respectively.

Metals, ceramics, and composites are generally applied to dental restorations and implants. The effect of friction has an important role to play in the mechanical function of dental devices. Friction between dental materials and bone affects the micro-motion and consequently fixation [56], similar to the fixation of artificial joints. Friction in fixed orthodontic appliance systems has been known to most clinicians to be harmful to tooth movement. Friction between brackets with different materials such as stainless steel etc. slid against various archwires was measured by Tecco et al. [57] and Fidalgo [58], with considerable differences between different designs and materials.

4.1.6 Catheter

Catheters and guidewires are widely used for medical diagnoses and interventions by inserting into a body cavity, duct, or vessel in order to allow drainage, administration of fluids or gases, or access by surgical instruments. There are numerous examples such as wound drains, endotracheal tubes; trochars; catheters; dilators; guide wires; angioplasty balloons; vascular, biliary and urethral stents; patches; filters; hypodermic or suture needles; and electrical pacemaker leads. Friction arising from this process directly results in shear stress that may damage the natural tissue and affect comfort, but also may influence the ease of insertion and manipulation in computer assisted surgery [59]. Various materials, particularly with coatings, have been developed over many decades to reduce friction [60].

Both *in vitro* and *in vivo* animal models were used to measure friction by Nickel et al. [61] and Khoury et al. [62] for different urinary catheter materials and coatings. *In vitro* measurement of static and kinetic friction coefficient of a catheter surface was performed by Kazmierska et al. [63]. Contacts between different counter-faces (polymers, tissue) and various types of tubes under wet conditions were simulated in order to mimic *in vivo* process. Low friction was found for super-hydrophilic biomaterials on tissue and a hydrophobic counter-face, while slightly hydrophobic biomaterials showed higher friction in both cases. More hydrophobic biomaterials gave low friction on tissue but high on hydrophobic polymer. The smoothest friction characteristic was achieved in all cases on tissue counter-faces. The static coefficients of friction of catheters on bladder mucosa counter-faces were measured as 0.15 for vinyl and siliconised latex catheters and 0.05 for all-silicone. Hydrogel coated catheters exhibited the lowest static and kinetic friction factors. The use of a hydrophilic-coated catheter during transradial cardiac catheterization was also shown to be associated with a low incidence of radial artery spasm [64].

4.2 Outside the body

4.2.1 Skin

Skin is the largest organ in the human body. Friction studies on skin can provide valuable insight into how the skin interacts with other surfaces and changes under various conditions including age and health, chemical treatments using lotions and moisturizers. Friction between skin and cloth may affect how we feel, and slips when entering or leaving a bath may be a serious hazard particular for the elderly [65]. Blister and pressure ulcer formation are also closely related to skin friction.

As an external surface itself, it is convenient, relatively easy and non-invasive to measure skin friction *in vivo* quantitatively. Friction studies on skin have been carried out comprehensively. Most tests have been performed *in vivo*, with a few *in vitro* and on animal skins. Friction measurements have adopted two basic designs: a probe moved across the skin in a linear fashion or a rotating probe in contact with the skin surface. Specific designs for friction measurements have been comprehensively reviewed by Sivamani and Maibach [66] and Derler and Gerhardt [67].

Coefficients of friction of skin at different anatomical sites, against various counterfaces and in the presence of various chemicals and under different actions have been summarized by Sivamani and Maibach [66]. It has been generally noted that skin friction depends on anatomical site and skin hydration as well as the design of the measuring instrument and the counterface geometry and material. However, no significant differences have been found with regard to gender or race [68]. The effect of age on skin friction may be linked to the increased sunlight exposure which can affect the skin structure and, therefore, alter the friction properties of skin. However, no significant differences of friction have been found with regard to age [69]. The coefficients of friction in the normal untreated skin generally range from 0.2 to 0.5, and under some conditions can reach as high as 2. Representative values of the coefficient of friction for normal dry skin from different anatomical sites range from 0.40 (leg, hand (dorsal)), 0.49 (forehead), 0.68 (hand (palm)), 0.81 (finger), and 1.20 (foot (sole)) [70]. Despite a complex underlying tribological mechanism, skin hydration appears to be the most important factor, followed by the influences of surface and material properties of the contacting materials. Friction increases with skin hydration and decreases for dried skin. However, the presence of a slippery layer of water may reduce friction through hydrodynamic action. Chemical treatments influence skin hydration level and affect the friction coefficient.

High friction can result in skin blisters, commonly found in active populations. Friction blisters not only create localized discomfort but also potentially serious secondary complications such as cellulitis and sepsis. Most research on friction blisters has been carried out from the military because of the nature of the physical activity involved in this field, as well as in the field of sports medicine. Prolonged pressure on the skin surface such as on the heel and associated friction and shear is related to the pathophysiology of pressure ulcers [71].

The effect of friction on touching, sensing and perception has received significant attention recently in a number of studies. Tactile sensation is usually assessed through the combination of friction measurements with objective correlation with other physiological parameters [72–75]. The underlying mechano-transduction in the skin sensing has been discussed by Zahouani et al. [76]. The mechanical skin sensation in humans can detect and differentiate many mechanical stimuli from the surrounding environment, for example vibration, texture, pinching, etc. These mechanical stimuli may exert deformations on the nerve ending in the skin with specialized sensitive receptors (mechanoreceptors). Friction affects the skin deformation and hence is directly related to this mechano-transduction process.

Friction of human hair has long been studied. The differential friction effect has been observed for many years when sliding direction along the hair is changed. A differential coefficient of friction of 0.16 was measured by Bhushan et al. [77] between polyurethane sheet sliding against Caucasian hair. Shaving and corresponding technologies are also closely related to friction [70]. One of the notable developments is the low friction PTFE coatings which are widely applied on the cutting flanks of the built-in blades in disposable razors.

4.2.2 Slips

Friction between feet/shoes and the floor influences the propensity of pedestrians to slip and fall. Clarke et al. [78] defined a pedestrian slip as occurring when "the required friction exceeds the friction provided from shoe-surface contact and the person fails to alter their gait (motion) accordingly". One of the common sources for causing unintentional slips and falls is bathtubs and showers. Friction studies have placed a major role on modern footwear development. Coefficient of friction provides a good indication of the slip resistance between footwear and a surface. During a gait cycle, the coefficient of friction required by a person can be described as the ratio of the horizontal to the vertical component which can be measured from a force platform. The biomechanics of slips were studied by Redfern [79]. The maximum coefficient of friction required occurs at the heel impact phase and the propulsion phase. Generally, the lower the friction between shoe-floor surfaces is, the more likely slips occur. Fiction coefficients less than 0.24, greater than 0.36 and between 0.24 and 0.36 have been suggested to correspond to danger, safe, and marginal risk (http://www.tribology.group.shef.ac.uk/

research/research_projects_banana.html). The presence of a banana skin may increase the slip risk, particularly when it becomes old, soggy, and brown. However it is difficult to use the ground reaction data alone to predict the likelihood of pedestrian slip due to the subjective nature of human walking and testing. Examples of uncertainties include large natural variability between individual humans (age, weight, body shape etc.) and extrinsic factors (surface and footwear characteristics). The walking velocity, as well as a person's ability to adapt their gait to particular footwear and surface conditions, are also important.

A number of mechanical testing devices have been used in the assessment of surface slip resistance in the form of friction coefficients, as summarized by Clarke et al. [78]. Chang et al. [80] outlined the detailed requirements in terms of the normal force build-up rate, the normal pressure and sliding velocity at the interface and the time of contact prior to and during the friction measurement. Although these mechanical devices can provide useful and re-producible data, inherent complexities in mechanically simulating subjective human gait make the validation of test devices difficult. Nevertheless, important parameters include shoe design, material, ground surfaces and conditions as well as individual gait characteristics. Table 3 summarizes typical representative coefficients of friction in shoe-floor contacts.

Table 3 Coefficients of friction measured between a PVC sole with a smooth PVC heel under various floor conditions [81].

Floor	Conditions	Coefficients of friction
Vinyl composite tile	Dry	1.12
Carpet	Dry	1.43
Vinyl composite tile	Wet	0.64
Carpet	Wet	0.80
Vinyl composite tile	Soapy	0.16
Carpet	Soapy	0.46

5 Summary

In general, friction measurements are relatively easier to conduct than lubrication and wear studies, and therefore have been carried out widely in tribological investigations of biological systems. Friction plays an important role in the normal function and potential disease development of a number of human organs as well as the development of diagnostic and interventional medical devices. Friction is usually measured in simple apparatus using small samples *in vitro*. The importance of these simple laboratory experiments in revealing basic biotribological mechanisms is widely recognized and is particularly useful for the purpose of comparative studies. However, this can result in a wide range of values of coefficient of friction reported, even when a similar tissue is considered. It is now recognised that good simulation of the *in vivo* situation is essential if laboratory observations are to be representative of *in vivo* performance and in design studies and the pre-clinical evaluation and screening of implanted products. It is increasingly clear that the physiological conditions should be replicated as fully as possible in order to provide meaningful indications of *in vivo* performance. Although friction studies generally provide valuable information in terms of friction coefficients, the underlying tribological mechanism remains unclear in most of the organs reviewed in this paper. It is also clear that friction measurements in terms of magnitude alone are often insufficient since a higher value may be associated with fluid-film lubrication while a lower value may be a result of some of the remarkable forms of boundary lubrication adopted by nature.

Different lubrication mechanisms have developed to control friction in different organs and tissues. However, for the majority of soft tissues, such as articular cartilage, cornea, pleura, fad pat etc., where sliding is important, it is intriguing to recognise basic similarities between the tissue compositions (biphasic in terms of solid and fluid phases) and the mechanisms of lubrication and friction adopted by these tissues engaged in different functions. Similarly, bio-lubricants associated with different biological tissues and organs have similar constituents including synovial mucin, hyaluronic acid, proteoglycans, glycoproteins (lubricin) and lipids (dipalmitoyl phosphatidylcholine, DPPC). Most interfaces in biological systems operate in a mixed lubrication regime, as do many engineering systems, with the ability to accommodate boundary, fluid film or a mixed lubrication regime to meet functional needs. Many of the basic mechanisms of boundary and fluid film lubrication are operative at different anatomical sites. Under the conditions in favour of hydrodynamic lubrication, a fluid film lubrication regime is responsible for low friction. Under conditions when contacts take place, the biphasic nature of the soft tissues takes the advantage of the fluid pressurization and the reduction in the load carrying proportion by the solid phase under external loading, so that the friction remains low for a considerably long period of time. Even when either the fluid-film or biphasic lubrication mechanism ceases to operate, the effective boundary lubrication mechanism comes into play and keeps friction adequately low. It is such a remarkable combination of different lubrication mechanisms that are responsible for the low friction observed in a majority of the soft biological tissues under a wide range of operating conditions.

Differences of the bio-friction in living biological tissues from the mechanical counterpart in engineering systems should be recognized. Natural biological tissues such as articular cartilage have self-regenerating ability, including friction and lubrication. The role of sliding motion and frictional shear stress has been shown to be important for regenerating functional extra-cellular matrix of articular cartilage and lubricating constituents (lubricin) on the surface in an *in vitro* set-up [82]. Similar regenerating mechanism may be expected for natural articular cartilage under *in vivo* conditions. For hard biological tissues such as teeth, self-repair or self-regeneration in terms of tribological properties is also expected to be important. Zheng et al. [83] have shown that the nanomechanical and microtribological properties of the acid-eroded enamel surface were significantly enhanced by remineralization in artificial saliva. However, the loss of the hardness and Young's modulus of enamel surface by acid erosion could not be fully recovered after *in vitro* remineralization. The understanding of the self-regenerating mechanism, including tribology, of biological tissues is important for not only understanding how our natural systems work and diseases may develop but also providing design guidance for developing effective tissue engineering approaches.

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