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Perspectives on the future of skin–prosthetic interactions

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Skin is filled with a rich network of sensory neurons, and consequently skin interaction with, and perception of the external environment, is a continual phenomenon experienced by all people in society. For medical devices such as prosthetics, the mechanical interaction between skin and the device is an essential aspect for device function, but it must be minimized to ensure continued skin health, which is inherently intertwined with biological processes. Numerous factors influence the interface properties of skin. For example, hydration, which influences skin morphology and composition, in turn influencing skin mechanics that may lead to tissue inflammation and skin injury. Here, we review the current state-of-the-art in skin medical device tribology with a particular focus on skin–prosthetic interactions. We split this article into traditional approaches to skin friction and a biology-first approach to skin friction, with a paradigm shift of skin as an engineered material. The field of tribology has historically been an interdisciplinary field comprising engineers, chemists and physicists, but future developments are needed in skin biology to drive meaningful change. We envisage that the integration of both clinical and biological perspectives will drive future innovations towards improved medical device interactions with the skin, when paired with engineering perspectives.

1. Introduction

Skin plays an important role in how we perceive and interface with our environment [1], ranging from heat and moisture management to tactile or pain stimulus detection. Critically, skin and its appendages provide a barrier to abrasion, infection or UV damage and protect underlying soft tissues [2]. Previous reviews have focused on the role of biotribology in everyday life [3], fingerpad and product interactions [4], artificial joints and dental implants [5]. However, the specific role that skin plays in medical device interactions has received less attention than articulating artificial joints such as hip or knee replacements. In particular, the impact that system properties have on the developed skin friction is an active area of research [6]. Much of the literature in skin biotribology takes an engineering-first approach which neglects the fact that skin is a dynamic and adaptive biological material. Additionally, the skin's response to loading will be biological in nature, and inflammation and skin dysregulation may onset before the visible rupturing of skin. Medical devices provide substantial benefits to users but have some major limitations, where much attention has been directed towards the manufactured medical device and less afforded to the device–skin interface. In this review, an outline of current research into skin tribology with medical devices will be presented, with a particular focus on prosthetics and an overview of the interdisciplinary challenges present in prosthetic socket design.

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2. The role of friction on skin

Skin responds in two very distinct manners to friction depending on loading modality, environmental conditions and anatomical location. Friction against the skin's surface can elicit pleasurable sensations with the most pleasant responses reported at speeds of 3 cm s^{-1} and normal forces of 0.2 N [7]. In contrast, higher pressures lead to skin irritation—this effect is exacerbated by the presence of shear stresses [8,9]. While the exact mechanisms that lead to skin breakdown are still being investigated, it is generally accepted that prolonged pressure and the addition of shear lead to skin breakdown. Klaassen summarizes two distinct mechanisms which lead to tissue inflammation, degradation and injury onset [10]. Firstly, combinations of pressure and shear lead to occlusion of blood vessels, build-up of waste products within the skin coupled with restriction of oxygen and nutrients provided by the vasculature leading to tissue damage over a longer time course (few hours). Secondly, high strains on individual cells lead to programmed cell death (apoptosis) or spontaneous cell death (necrosis), resulting from events such as rupture of the cell membrane, leading to subsequent degradation of the tissue over a shorter time course (few minutes) [10]. In parallel, inflammation caused by these two mechanisms leads to an increase in inflammatory cytokines. Research in type I pressure ulcers, so before skin breakdown, has shown elevated IL-1α, IL-1RA, IL-8, G-CSF and an increased ratio of IL-1α/IL-1RA in ulcer sites compared with patient-matched control skin [11].

Onset of skin irritation and injury can be exacerbated by high friction and shear stress on the skin due to the presence of high deviatoric stresses, which cause large deformations in soft tissues [12]. Developed skin friction is influenced by intrinsic factors such as sex, anatomical location and skin hydration status [13]. It is important that future skin research considers the varying needs of diverse populations, in terms of age, skin tone and sex [14]. Previous studies have reported that eumelanin-rich skin shows different structures and cutaneous responses to inflammation and scarring [15,16], meaning device–skin interactions pose larger risks for some. However, dermatological literature on darker skin phototypes is still limited. The majority of research studies are performed in Fitzpatrick scale I–II skin, making it harder for researchers and designers to build equitable products that cater to different needs [14]. Sex is also another consideration. Women's monthly hormonal cycles, or hormonal changes during perimenopause and menopause, are known to affect the skin, impacting skin thickness and wound healing capacity among other factors [17,18]. However, most tribological studies do not take this into account or attempt to link skin mechanics with these hormonal changes.

In addition to these intrinsic factors, extrinsic biophysical factors such as environmental temperature and relative humidity [19] and contacting material properties [13,20] also alter skin friction. These intrinsic and extrinsic factors influence the biological function of the skin and in turn alter its mechanical properties, subjecting the skin to greater deformation while also impairing repair and remodelling mechanisms essential for barrier reformation [2].

Despite advances in prosthetic devices, skin injuries remain an unmet clinical need for lower limb amputees. Pressure has been reported in several studies in the prosthetic socket, with direct pressures measured at 34–149 kPa during standing and 58–342 kPa during walking [21–25]. However, owing to the largely incompressible nature of soft tissue, these high peak pressures recorded on the surface give little indication of underlying tissue health [26]. Instead, shear stress may give a better indication to tissue health in a prosthetic socket. An amputee's residual limb is subjected to high levels of friction and shear to keep the prosthetic in contact with the skin and transfer the body load efficiently. Lower limb amputees using suspension systems require friction for device functionality, despite friction's role in skin breakdown. Socket shear stresses have been measured at 27–134 kPa and increase the likelihood of developing skin injuries [21,23]. Consequently, reports of skin problems in lower limb amputees range from 34% to as high as 75% [27–31], suffering a disproportionately higher skin injury rate (65%) when compared with the general population [32].

3. Traditional approaches to skin–medical device interactions

Medical devices, such as prosthetic limbs, personal protective equipment (PPE), non-invasive ventilation masks, splints, braces and bandages/medical adhesive tapes to secure the devices, can be attributed to one-third of hospital-acquired pressure ulcers [33,34]. Several groups have outlined the importance of regular skin assessments at areas of medical device contact and suggest a need for interdisciplinary medical teams and engagement with medical device designers to reduce risk of skin injury [35,36]. Skin friction in a prosthetic socket is a complicated and multifactorial problem consisting of an interplay between contact mechanics and biological feedback. Approaches to characterize and investigate skin friction highlight the importance of system properties [6] (figure 1). In the case of the residual limb, this interfaces with the prosthesis through a sock and silicone liner; this interaction helps keep the prosthetic device in place and allows load transfer to enable the user to walk.

3.1. Engineered material

Material properties of the countersurface ('Engineered Material' in figure 1) are known to influence friction intensity. In particular, higher friction coefficients are recorded in prosthetic socks made from wool and nylon due to coarse fabric weaves and protruding fibres, which lead to microscopic skin trauma [37]. Properties of contacting engineered materials also have effects on reducing friction, where Klaasen *et al.* found that engineered surfaces with a 4 µm surface roughness and higher stiffness (0.91 MPa) are more beneficial in scenarios where static friction needs to be minimized, and conversely softer materials (0.45 MPa) would ensure sticking (high static friction) and prevent dynamic sliding [38]. Altering interface friction in response to altering user requirements and biophysical conditions could help improve device efficacy. Efforts to reduce interface pressures by using softer contacting materials in hospital beds have shown limited efficacy in reducing prevalence of pressure

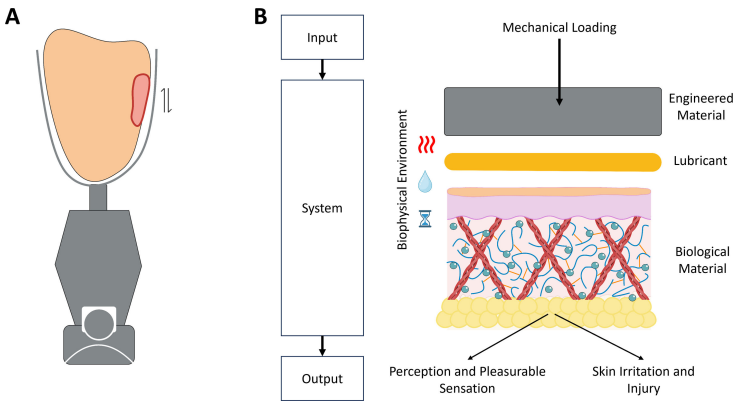

Figure 1. Skin tribology system in a prosthetic socket. (A) Schematic of shear present at skin surface of prosthetic socket. (B) Schematic demonstrating the crosstalk of engineering and biology that leads to skin discomfort and injury. All diagrams created in BioRender and Inkscape.

ulcers [39]. Oomens *et al.*'s computational analysis of a seated buttock showed that reducing the stiffness of the engineered material by a factor of 16 has minimal effects on the internal strains in the deep tissue [26]. However, Boyle *et al.* find that the addition of lateral pressure counterintuitively reduces internal deformations, providing a potential solution to ulcer risk patients in wheelchairs or bed-bound individuals [12]. Tanriverdi *et al.* trial the use of an adaptive prosthetic socket to account for volume fluctuations in residual limbs and find improved self-reported residual limb health [40]. While reducing interface pressures have yielded inconsistent results, Reuvekamp *et al.* provide a comprehensive review of future research directions and desirable outcome measures for functional surface engineering to control the microclimate of the skin–medical device interface [41]. The ideal microclimate for healthy skin is defined as 40–65% relative humidity, skin hydration levels of 10–20% and skin temperature at 16–33°C [41]. Socket microclimates have been reported at 80–90% relative humidity [42]; Reuvekamp *et al.* hypothesize the use of surface engineering techniques, such as patterned and porous surfaces to harvest and transport unwanted moisture, could alleviate these negative conditions [41].

3.2. Lubricant

Skin friction is defined by a non-interacting two-term equation [43] and is thought to be more strongly influenced by surface adhesion and surface chemistry [44]. Sebum is a fluid film on the surface of the skin and is a mixture of lipids released by sebaceous glands, located near hair follicles, that spreads across the surface of the skin [45]. Sebum is believed to play a role in UV protection, antibacterial capacity and have a function in inflammatory responses [46]. It has also been suggested that sebum acts as natural lubrication for hair follicles [47] and surrounding skin [46]. Preliminary research into the effects of sebum on skin friction has indicated weak correlations between lipid content and friction [48–50]. In sebum-poor areas such as the abdomen, the sebum thickness was less than 0.5 µm and the coefficient of friction was measured at 0.12 [51], while the forehead, where sebum thickness was greater than 4 µm, measured 0.34 [45]. While there is little correlation between poor and rich sebum areas, a deficiency of sebum does have a dramatic effect with a complete loss of surface sebum leading to a threefold reduction in the recorded coefficient of friction (0.3 versus 0.9) [48].

We (M.M.) previously argued that contributions of adhesion and deformation to skin friction are load dependent. The adhesive contribution to the coefficient of friction reduces with increasing load, while the deformation contribution to skin friction increases with increasing load [6]. Influencing the shear strength of the interface by addition of a lubricant can also reduce skin friction. Investigations by Yap *et al.* into healthcare PPE highlight optimal mixtures as 20 wt% beeswax, 40 wt% olive oil and 40 wt% mineral oil, which led to an 87% reduction in shear [52]. However, the complex nature of how lubricants interact with skin and alter the contact's system properties mean that emollients and creams should be avoided due to their secondary effects of altering tissue mechanics [53]. In the case of skin–prosthetic interactions, friction is a design requirement in non-osseo integrated designs. Reducing friction at the prosthetic–skin interface would lead to reductions in functionality, secondary pain in posture resulting from misalignment of the prosthetic and greater risk of falls from device misalignment.

3.3. Biophysical environment

Skin mechanics are influenced by the biophysical environment and as such must be controlled due to the relationship between soft skin mechanics, shear and resulting inflammation. Boyle *et al.* also highlight the interplay of skin mechanics on excessive deformations and friction in ulcer aetiology, which they term 'deformation ulcers' [12]. Skin tribology studies have been conducted in healthcare workers evaluating the use of PPE on soft tissue health. These revealed that short-term changes such as altered temperature and humidity levels greatly affect skin behaviour as seen when using facial PPE for prolonged periods of time [52–54]. Biophysical environments with elevated humidity also act to soften the skin, causing skin to conform more to the mask, increasing contact area and the associated adhesive forces [55] and leading to reduced tissue tolerance [56].

Occlusive environments are a pertinent problem with non-invasive ventilator masks [57–59] and incontinence products [60,61], with the effects being particularly pronounced in paediatric patients [58]. The use of accurate three-dimensional facial scans allows the creation of customizable masks, catering to diverse sizes and facial shapes [57]. Users of absorbent incontinence

products, such as back sheets, have increased risk of incontinence-associated dermatitis (IAD); nursing home residents have IAD incidence rates of 5.5% [60], while critical care patients have 29–36% [62] making it a prevalent issue. The TENA SmartCare Change Indicator tracks urine saturation in incontinency pads, reducing the time skin spends in contact with urine under pressure by indicating when a pad needs to be changed. When tested in a large-scale clinical study, it improved care efficiency and sleep quality of care home residents [63]. While the use of technology to help with incontinency is still new and developing, it is an example of how technology and biological understanding can be synergized to improve skin health.

Efforts to reduce skin injuries from prosthetic use have been focused on engineering-first solutions to reduce pressure and shear at vulnerable locations. A recent review by Zhang *et al.* suggests that skin friction is often required to maintain functionality of the medical device [64]. However, shear negatively impacts soft tissue health with the environment created by a prosthetic socket impacting skin mechanics, increasing vulnerability to skin irritation and injury. Some promising advancements in adaptive sockets that can grow with the user, or engineered surfaces and control measures that can dynamically modify the microclimate, can mitigate exacerbating factors that worsen skin health. However, a major part of the interface remains unaltered and under-investigated. Here, we explore the potential to vary skin properties in the prosthetic environment as a novel approach in skin tribology research.

4. Biology-first approach to the medical device interface

4.1. Skin biology

Skin is an inhomogeneous adaptive material with three main functional layers: the epidermis which interfaces with the external environment, the underlying dermis which provides mechanical support, houses functional structures like glands, hair follicles and vasculature and the hypodermis (fat layer) which dissipates mechanical force (figure 2) [47,65]. Skin principally acts as a two-way barrier providing mechanical protection from the external environment and retaining water within the body [65]. Skin structure and barrier function are established and maintained by a complex interplay of biological materials such as lipids and continued expression of keratins, a structural intermediate filament found in cells within the skin epidermis [66] (figure 2).

4.2. Dermis

The dermis is approximately 2–5 mm thick depending on body site [67] and the predominant cell type is fibroblasts. Fibroblasts play a role in tissue architecture [68], wound repair [2] and orchestrate overlying epidermal development [69,70]. The dermis can be split into two distinct regions. The papillary dermis lies directly beneath the epidermis and comprises approximately 10% of the dermal thickness [71]. Here, fibroblast cell density is highest and elastin and collagen content is lower [47]. Below lies the remaining 90% of the dermis known as the reticular dermis. This layer is characterized by a higher abundance of collagen fibres and lower cell density [47]. Fibroblasts are traditionally defined as being derived from the reticular or papillary dermis; however, single-cell sequencing of human skin has highlighted four distinct fibroblast subpopulations [72], while spatial transcriptomics have outlined six fibroblast subpopulations in healthy skin [73].

4.3. Epidermis

In contrast to the dermis, the epidermis is avascular and therefore relies on diffusion through the barrier and between cells for oxygen, nutrients and water (figure 3). This layer is stratified squamous epithelium consisting mainly (approx. 90%) of keratinocytes, with a small subpopulation of specialized cells [47]. Cells delaminate (exit the basal layer) then undergo differentiation coupled with stratification to move up through the epidermis to reach the stratum corneum.

The stratum corneum is the most superficial layer of the human epidermis (figure 3) which acts as a barrier and interacts with the external environment. Cells in the stratum corneum are sloughed off approximately 28 days after they have delaminated from the basal layer [47]. The stratum corneum comprises a lipid matrix aggregated with protein-rich anucleated keratinocytes termed corneocytes (figure 3). Vyumvuhore *et al.* found that at 9% tensile extension of excised human abdominal stratum corneum, intercellular lipids become more disorganized (gauche transition) [74] and *in vivo* rubbing of skin leads to reductions in lipid content in the superficial stratum corneum [75,76]. Maintaining a healthy barrier is pivotal to preventing skin dermatoses in lower limb amputees. However, recent work has shown prosthesis use leads to reduced skin health [76].

4.4. Skin as an engineering material

Skin is a heterogenous composite structure, which is pre-tensioned *in vivo*, exhibits time-dependent behaviour, hyperelastic strain behaviour and direction-dependent mechanics. In addition, all of these factors vary depending on anatomical location, current general health status, skin pathologies, biophysical environment and demographic of individual.

Skin layers contribute uniquely to bulk tissue mechanics. The stratum corneum has the highest elastic modulus followed by the viable epidermis and the more compliant dermis [77–81]. The stratum corneum plays an important role in the overall mechanical response of the skin; however, a large range of elastic moduli of all skin layers has been recorded in the literature [78,79]. This has been largely attributed to extraction methods for measuring isolated stratum corneum and the hydration state of the tissue. The effective modulus of the stratum corneum has been shown to vary between wet and dry conditions, being over

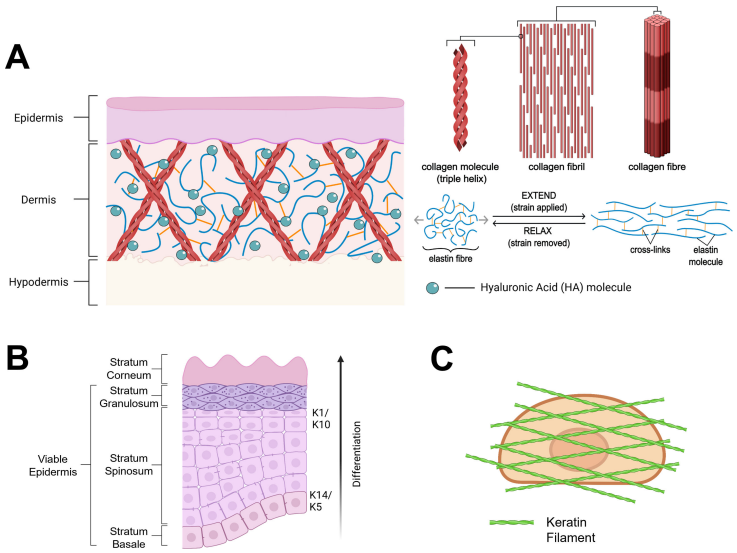

Figure 2. Skin organization and the load-bearing structures present. (A) Whole skin diagram, with collagen and elastin architecture. (B) Stratification and keratin distribution in the epidermis. (C) Keratin reinforcement of the cytoskeleton in individual keratinocytes. All diagrams created in BioRender and Inkscape.

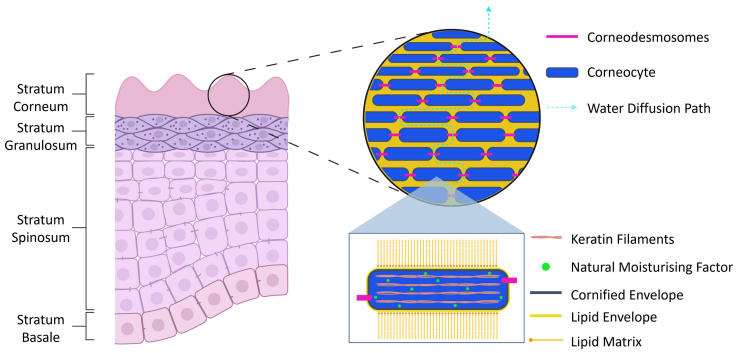

Figure 3. Human skin epidermal architecture. All diagrams created in BioRender.

600 times stiffer in the dry condition compared with hydrated condition (4000 versus 6 MPa) [82] which further implicates skin hydration and microclimate as an important player in stratum corneum health and mechanics.

Mechanics of the viable epidermis are driven by keratinocyte mechanics [83] where the main structural proteins are keratins. Isolated keratinocytes have an indentation stiffnesses of 1–3 kPa [84], but knockdown of *in vitro* keratin 5 expression led to keratinocytes with half the stiffness of control keratinocytes (412 versus 752 Pa) [85]. Within the dermis, the mechanics are dominated by collagen, produced by resident fibroblasts, which themselves have mechanics of 0.9 kPa [86]. Elastic energy storage occurs in the flexible regions of the collagen fibrils and stress transfer occurs through rigid sections of the molecule [87]. Mechanical load is bore through collagen, which is 10-fold stiffer than elastin [88]. The mechanics of the dermis reduces in stiffness by a factor of 3.5 as a result of age-induced disruption of collagen fibres [87].

Skin is a dynamic biological system and its mechanics can change because of medical device interaction. For example, after long-term prosthetic use, total lipid contents (ceramide and fatty acids) are significantly ($p < 0.05$) lower in the stratum corneum of residual limb skin compared with the intact limb, suggesting impaired barrier function after interaction with the medical device [76]. Natural moisturizing factor is also reduced at the skin's surface due to prosthesis use, which in turn leads to higher water content near the skin's surface. These biological changes have large-scale effects on soft tissue mechanics, causing a softening of skin by a factor of 3 and an increase of skin friction coefficient from 0.6 to 0.7 [76]. In turn, this leads to higher levels of inflammatory cytokines IL-8, IL-2 and IL-1β in residual limbs compared with intact limbs [76]. In a cycle of events, the augmented tissue mechanics (softer skin and increased friction coefficient) after prosthesis use makes skin more susceptible to deformation-based injuries caused by prosthesis use. This is reflected in high skin injury statistics for prosthesis users and highlights that the role of the tissue in skin–medical device interaction plays just as an important role as the device itself.

5. Augmenting skin–medical device interaction by changing the skin

As skin–prosthetic interactions depend on properties of both the device and the skin, research to date has largely focused on altering medical device interface properties to augment the interaction. However, we argue here that another approach should be considered in parallel with innovative engineering solutions in future research and that is to alter the skin instead [89]. Skin is the largest organ on the body and as such exhibits large regional heterogeneity enabling it to fulfil its function according to location. For example, skin on the soles of the feet is 1.6-fold stiffer in indentation and 2.4-fold stiffer under shear loading [78].

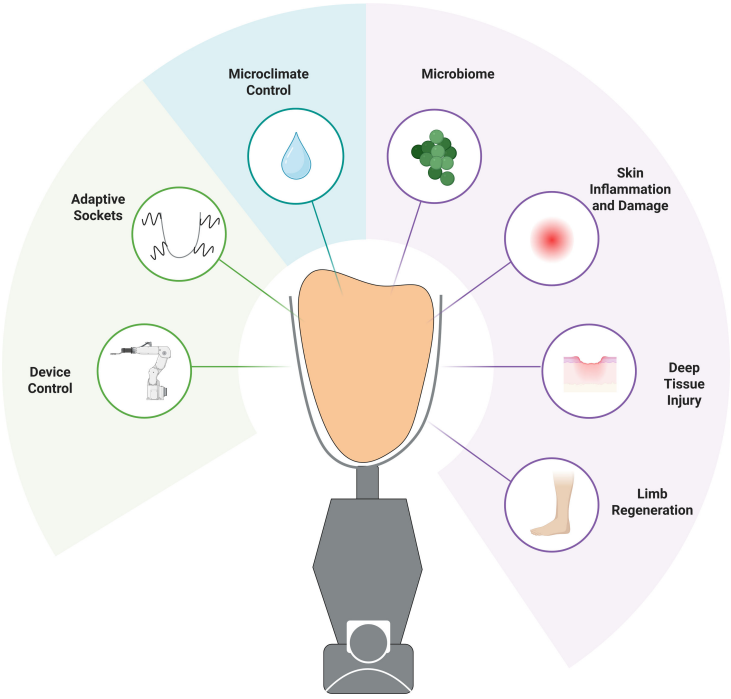

Figure 4. Interdisciplinary challenges in prosthetic provision.

The composition of palmoplantar skin protects it from a diverse range of adverse mechanical stimuli [78]. In contrast to the features present on the plantar heel, skin on a lower limb amputee's residuum lacks unique keratin expression and morphological features to confer improved load bearing, resulting in high skin injury prevalence. Re-engineering of non-load-bearing skin, such as that on the residuum, towards a load-bearing phenotype that mimics plantar skin mechanics has been predicted to reduce shear strains by 47%, which would consequently reduce failure by biological fatigue [89]. Altering the mechanics of residual limb skin would allow keratinocytes in the epidermis to resist higher stresses without rupture, while increased stiffness in the dermis would allow greater resistance to distortion under the shear loading. One approach for augmenting the skin is using cyanoacrylate liquid skin protectant treatment which leads to an increase in the stiffness of the skin and an approximately 40% decrease of the coefficient of friction between the treated skin and contacting material [90]. These changes, however, are not permanent and ultimately could reduce effectiveness of a prosthesis.

Another approach is to graft plantar, load-bearing skin, directly to the residual limb which would enable skin to be weight bearing. Fuchs *et al.* demonstrated that when human plantar skin-derived microtissue columns were transplanted to the trunk of a mouse, they retained their plantar skin phenotype [91]. However, the key question is how can you re-engineer non-plantar skin to a plantar phenotype. For this, we can learn from nature as it is well accepted that skin fibroblasts orchestrate both dermal and epidermal identity during skin development [92]. Yamaguchi *et al.* first investigated the effects of co-culturing plantar fibroblasts with non-plantar keratinocytes, showing that the plantar fibroblasts could induce expression of a palmoplantar-specific keratin (KRT9) in non-plantar keratinocytes in as little as 30 min [93]. When plantar and scalp fibroblasts are compared in an *in vitro* three-dimensional skin equivalent model, changes in the epidermis are induced including increases in epidermal thickness (80 versus 60 µm) and increased plantar-specific keratin expression (1.5-fold versus 1.0-fold) for plantar and scalp fibroblasts, respectively [94]. Collectively, these data suggest that plantar fibroblasts have the capacity to reprogramme non-plantar keratinocytes to a plantar, and therefore theoretically load-bearing, phenotype.

The idea that plantar fibroblasts could reprogramme skin on the residuum to a plantar, load-bearing phenotype was recently tested in a phase 1 clinical trial with 39 lower limb amputees [94]. The study investigated the therapeutic potential of injection three times per week of 10 million isolated and cultured human plantar fibroblasts. The authors found that 17 months of plantar fibroblast injection leads to increased elastin content, increases in KRT9 expression and increases in epidermal thickness [94], which suggest a transition towards a plantar identity. The authors quantified skin 'firmness', with a Durometer (Buffalo Grove, IL, USA), confirming previous studies [78] that palm skin is substantially firmer than non-palmoplantar skin. However, they found that injection of fibroblasts, regardless of the anatomical location from which they were harvested (plantar skin or scalp), led to increases in skin 'firmness' [94]. This suggests that using plantar fibroblasts to re-engineer skin on the residual limb has merit, although more comprehensive studies need to be conducted looking at their role in augmenting skin mechanics.

6. Conclusion

Skin is a dynamic regenerative organ that interacts mechanically with the external environment. Its main function is as a barrier to mechanical damage; however, it also acts to mechanically interact and sense the environment. Skin tribology is inherently intertwined with biological processes. Numerous factors influence the interface properties of skin, namely, hydration influences morphology and composition which in turn influence mechanics leading to inflammation and skin injury. Highly orchestrated

cellular processes allow conservation of the epidermal barrier through spatial differentiation. Skin morphology and composition vary based on anatomical location, with the most pronounced contrast occurring between loading skin (palmoplantar) and non-loading skin (non-palmoplantar). Recent characterization of skin covering an amputee's limb shows substantial adaptation occurs with wearing of the prosthesis that does not constitute greater resistance to load [76] and instead results in biological fatigue.

Characterizing skin of different pathologies and different mechanical environments is essential in soft tissue problems. It is important to improve the breadth of skin characterization to encompass groups that have been researched less, such as women in menopause. Additionally, moving away from studies focused solely on a single sex and/or genetic ancestry (usually males of European ancestry) is key to obtaining a more holistic and equitable view and understanding of skin mechanics. This research must be conducted with an integrated approach (figure 4) due to the interdependence on skin biology and skin mechanics. Throughout this piece, we have argued that a biology-first approach to skin tribology is required to complement the engineering-first approach currently used. We highlight the viewpoint of skin as an engineering material, which can open exciting possibilities and solutions to mechanical problems in healthcare.

Ethics. This work did not require ethical approval from a human subject or animal welfare committee.

Data accessibility. This article has no additional data.

Declaration of AI use. We have not used AI-assisted technologies in creating this article.

Authors' contributions. J.H.: conceptualization, funding acquisition, investigation, project administration, resources, visualization, writing—original draft, writing—review and editing; M.M.L.D.S.: visualization, writing—original draft, writing—review and editing; C.A.H.: supervision, writing—review and editing; M.M.: conceptualization, project administration, resources, supervision, visualization, writing—original draft, writing—review and editing.

All authors gave final approval for publication and agreed to be held accountable for the work performed therein.

Conflict of interest declaration. We declare we have no competing interests.

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