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On the way to total integration of prosthetic pylon with residuum

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Abstract

Two decades after introducing threaded titanium dental implants, Dr. Per-Ingvar Brånemark used a similar technique in the 1980s to pioneer the direct skeletal attachment (DSA) of limb prostheses. He and his colleagues used convincing clinical experience to overcome the skepticism of their peers, affording a new dimension of prosthetic rehabilitation to almost 100 individuals with amputation. As a result, more research has been initiated worldwide to move DSA to a level of greater safety, longevity, and reliability. This review highlights the trends and milestones in current DSA development. It also identifies ideas from previous studies in various fields that may be useful in future DSA development.

Keywords

amputation; arthroplasty; bone; direct skeletal attachment; infection; limb amputation; osseointegration; prosthetic rehabilitation; skin; transcutaneous devices

INTRODUCTION

As of 2008, about 100 individuals with amputation have had abutments for attaching leg prostheses implanted in their residuum [1]. Dr. Per-Ingvar Brånemark developed the technology for direct skeletal attachment (DSA) of limb prostheses after discovering that a firm bond exists between the titanium implant and the surrounding bone. He called the bond “osseointegration” [2]. Brånemark discovered osseointegration in an orthopedic model by implanting the titanium device into the medullary canal of an animal tibia [3], but the first translation of the research to human applications occurred in the field of dentistry [4]. Brånemark introduced his method for DSA of limb prostheses after almost 2 decades of continuous development and refinement of his successful method in dental implantology [5].

At the 2004 International Society for Prosthetics and Orthotics World Congress in Hong Kong, one of Brånemark’s patients demonstrated how his transfemoral prosthesis could be attached to and detached from the abutment implanted in his residuum. I was impressed by the demonstrator’s quality of ambulation and high level of satisfaction with and enthusiasm for DSA technology. However, the demonstrator pointed out the layer of pus between the abutment and skin of the residuum (Figure 1), which was an obvious indicator that an infection-free seal did not exist around the abutment. This observation prompted me to consider integrating the abutment with the surrounding skin. Thus, a project was commenced to explore the hypothesis that a porous implant would create the proper conditions for the skin to form an infection-free seal [6]. A safe skin-device interface for

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DSA is part of the broader problem of infection-free transcutaneous devices, and this article presents a review of different approaches to the problem.

Another aspect of DSA that will be presented and discussed here is the risk of loosening of the implant in the medullary canal and the existing approaches to addressing this problem.

CHALLENGE OF LONGEVITY FOR DIRECT SKELETAL ATTACHMENT

In 2007, approximately 1.7 million persons were living with limb loss in the United States [7]. The main cause of acquired limb loss is poor circulation in the limb as a result of arterial disease, with more than half of all amputations occurring among people with diabetes mellitus. Amputation of a limb may also occur after a traumatic event or for the treatment of bone cancer [7].

The U.S. Department of Defense reported that between September 2001 and January 12, 2009, 1,286 individuals underwent amputations, of whom 935 had major limb amputations during Operation Iraqi Freedom in Iraq, Operation Enduring Freedom in Afghanistan, and unrelated conflicts. Of the 1,286 total persons with amputation, 77 percent sustained their injury while in the Army, 19 percent while in the Marines, 2 percent while in the Navy, and 2 percent while in the Air Force [8]. Expanded military operations have decreased the mean age at which amputation occurs. The survival rate after severe gunshot and land mine trauma has concurrently increased significantly. The increased survival rate has been made possible by improved immediate medical care on the battlefield, fast transporting of the wounded soldiers to definitive care facilities, better instrumentation that allows advanced external fixation of the traumatized limbs, and new surgical materials and techniques. As a consequence, the ratio of “number of amputees to number of death casualties” has increased to 30 percent from 7 percent during World War II and from 15 percent during the former Soviet Union’s 1979–1989 war in Afghanistan [9].

DSA of limb prostheses is a promising alternative to the traditional “socket-residuum” attachment. DSA can provide actual osseoperception, improving all locomotor activities of a patient, and eliminates the problems associated with donning and using a socket [10]. In light of the growing attention to neural prosthetics [11–12], DSA may provide in the future a platform for reliable transmission of signals from peripheral nerves to controllers outside the body [13]. However, prosthetics and rehabilitation professionals face the double challenges of rising number and rising life expectancy of amputation survivors. One must ensure that the skin-device interface is safe and sustainable and that the bone-device bond is sufficiently long-lasting. For these new challenges to be met, the existing or potential problems related to all components of the DSA technique and instrumentation should be carefully examined and addressed.

DIRECT SKELETAL ATTACHMENT OF LIMB PROSTHESIS: BRÅNEMARK’S METHOD

To eliminate the negative outcomes of the socket-residuum attachment, such as discomfort, pain, and secondary trauma [14], Brånemark and his colleagues introduced a method of attaching a prosthesis directly to the bone of the residuum [10,15]. Reasonable indications for DSA include a short residuum, an excessive volume of soft tissues, skin problems, and pain, which make the traditional residuum-socket attachment a hardly viable option. In upper-limb prosthetics, DSA can better restore grasp in patients with thumb amputations [16].

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The DSA technique is similar to that developed by Brånemark for dental implantology (Figure 2): a titanium threaded fixture is inserted into the bone remnant of the residuum, and the prosthesis is attached by an abutment. Albrektsson and colleagues reported that the threaded fixtures, manufactured of pure titanium and without any coating, were removed for various reasons from 18 patients after loading up to 90 months and demonstrated direct chemical bonding between the bone and titanium [17].

DSA requires a long-lasting bond between the implant and the hosting bone and an infection-free transcutaneous passage. These requirements are also critical in the other areas of implantation. Experts in arthroplasty (total joint replacement) have sought to improve the bone-device bond for decades [18–20]. A safe skin-device interface is of great importance for dialysis [21–23], trachoeostomy [24–26], and feeding and drug delivery in chronically ill patients [27–28]. Therefore, DSA may benefit from research and ideas from other fields.

SKIN-IMPLANT INTERFACE

The problem of infection in the device-skin interface remains unsolved; infection as an adverse effect is reported in a high percentage of cases [29–31]. Studies were undertaken to promote and increase the binding junction with the skin by testing different materials with various circumferential components [32]. Attempts have been made to enhance the interface between the implants and skin cells by creating open pores on the surface of the implant [33], making fiber metal composites [34], treating the surface of the implants with calcium phosphates [35–36], varying thermal conditions [37], and coating the metal with specific agents to decrease mechanical mismatch between the implant and skin or bone [38]. The interface has been evaluated by measuring adhesion, proliferation, and differentiation of cells [39–40]. Cell morphology and cytoskeleton have been evaluated with scanning or confocal microscopy by the synthesis of the proteins of the extracellular matrix and proteoglycans [41]. In vivo, biocompatibility has been evaluated histologically [42–43].

EFFECTS OF POROSITY

Porosity was considered a logical means for improving the integration of devices with hosting biological tissues like tubular bone [44–50], spine [51–53], skin [54–58], and tendon [59–61]. Empirically, the volumetric porosity of an implant should be at least 30 to 50 percent for surrounding tissues to grow inside. Increased porosity, however, significantly decreases the strength of the implant. New technologies combining powder metallurgy and polymer foam technologies demonstrated some progress in manufacturing titanium and tantalum foam with strength close to trabecular bone but not exceeding compact bone [62–63].

Porosity of percutaneous implants has been a point of interest for many researchers. Von Recum summarizes the areas of application of the percutaneous implants, including internal/external prosthetic devices, blood access devices, tissue access devices, body cavity access devices, and power and signal conduits [64]. He also describes the principal failure modes in the applications of the percutaneous devices, as identified by Lee et al. [65], Hall et al. [66], and Winter [67]. These modes are (1) extrusion due to *marsupialization*, (2) extrusion due to *permigration*, (3) extrusion due to *infection* and abscess formation, (4) extrusion due to *avulsion*, and (5) extrusion due to any combination of the just listed failure modes.

Marsupialization is shown in Figure 3(a), with the epidermis migrating and proliferating inward along the solid implant. Also, when a penetrating implant has no pores or pores with diameter less than 40 μm, a sinus tract is formed [66]. The use of a porous implant surface was proposed to avoid marsupialization and the sinus tract infection identified in a study with Dacron velour [68]. In that study in pigs, a percutaneous implant for attaching an

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artificial limb had an intramedullar stem with porous surface and Dacron velour was used at the soft tissue interface. The ingrowth of both bone and soft tissue was seen. However, the authors reported that velour was unable to maintain adequate epithelial adhesion to form an anatomical seal and a barrier to bacteria. The failure of the permigration mode (Figure 3(b)) in the use of Dacron velour has been confirmed by von Recum [64].

In 1970, Hahn and Palich reported positive results using porous titanium in orthopedic applications [69]. Clemow et al. have examined the interfacial shear properties of bone tissue growth into porous-coated Ti6A14V femoral implants as a function of the pore size of the porous surface [70]. Cylindrical implants with powders of three particle-size ranges (297, 420–500, and 595– 707 μm) were manufactured. After sintering, the fine, medium, and coarse powders resulted in pore sizes of approximately 75, 225, and 325 pm, respectively. The implants were inserted for 6 months into the femoral medullary canal of dogs. Push-out tests on the removed femurs revealed that the interfacial shear strength and stiffness of the bond decreased as pore diameter increased within the range of 175–325 μm.

The antimicrobial feature of titanium oxide (TiO_2) has been demonstrated and used to explain the low failure rate (0.07% per month) of hearing aids percutaneously anchored to bone with Brånemark's procedure [71]. The antimicrobial properties of TO_2, along with fibroblast adhesion to it, have been confirmed on Dacron fibers coated with a thin layer of TO_2 [72]. In a study on implanted biosensors, the anti-inflammatory properties of TO_2 were demonstrated with a small amount of TO_2 covering only 23 percent of the surface area [73].

The effectiveness of a barrier to infection in a skin-implant interface depends on the quality of the proliferation and attachment of keratinocytes to the surface of the implant. The attachment of the skin and bone cells to titanium surfaces is so strong because a very thin layer of titanium peroxy compounds is in contact with the living cells [10,74]. Surprisingly, Pendegrass and colleagues postulated that surfaces with a smooth topography of the titanium alloy implant at the point of epithelium-implant contact could increase attachment in vivo, producing an effective barrier against infection [75].

If we go back to the instance of pus between the abutment and skin (Figure 1), the failure to prevent chronic inflammation can explained by avulsion,* which is the other failure mode described earlier [64].

Avulsion is a major problem for limb DSA, since the mobility of skin reaches its maximum in the distal zone of a residuum. The average mobility in patients with excessive volume of soft tissues in that zone is even higher.

To minimize avulsion, the mobility of skin around the implant has to be significantly reduced. A surgical technique developed by Brånemark and his colleagues for this purpose includes removing the fat and thinning the surrounding skin to the thickness of a split-skin graft, allowing for skin adhesion to the bone end [16]. Devices for reducing avulsion have included a percutaneous bar with a flexible mesh collar, holes at the subcutaneous perimeter [76–77], and a collar made of a stainless steel spring or nylon hooks [78]. Animal studies with these devices produced promising results. However, the implants with a connecting collar are sensitive to its positioning relative to the derma and subcutaneous tissues and may not tolerate junction shifting when the distance from the bone to the skin-binding junction changes [78].

*A tearing away or forcible separation. Stedman's Medical Dictionary. 25th ed. Baltimore (MD): Williams & Wilkins; 1990. Avulsion; p. 159.

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Another approach was positioning of a bar with a porous flange in the dermal tissues immediately below the epithelium [79–80]. While it may reduce the mobility of skin in the plane parallel to the flange, the attachment to the solid bar still remains fragile, similar to the prior art.

In 2004, we initiated in vitro and in vivo studies on a totally porous pylon for DSA [81]. We suggested that resistance to detachment (avulsion [64]) will be increased because of the natural bond of the cells outside the pylon with the cells inside the pylon. This resistance is illustrated by a schematic in Figure 4, in which F_1 and F_2 are the minimal forces needed to detach a skin cell A from the pylon with the traditional "cell-to-wall" and a cell B with the "cell-to-wall-to-inner cell C" attachments, respectively. If a natural bond is permitted between the cells outside and inside the pylon through a pore in the pylon wall, an additional "cell-to-cell" attachment (adhesion) force contributes to the resistance to the cell's detachment, thus providing $F_2 > F_1$ [82].

A recent histopathology study demonstrated deep ingrowth of skin and bone throughout the novel pylon (i.e., skin and bone integrated pylon 2) [83], whose strength exceeds that of human bone [84].

BONE-IMPLANT INTERFACE

The literature tends to perceive the interface of the implanted fixture with the bone as a less serious problem than its interface with the skin [10,38,85]. We believe that this perception might be true in the short term. Long-term effects must be considered, since DSA is intended for lifelong use and present reports on the long-term applications of the DSA are still limited [86].

Current DSA technology uses threaded titanium fixtures that are screwed into the medullary canal of the residuum bone [5]. Bone remodeling has been observed inside the threads and on the surface of the metal fixture [86]. However, the procedure of screwing damages endosteum† integrity, resulting in endosteal absorption and consequent medullary canal widening, which may cause clinically unstable implants [87]. The medullary canal also widens naturally: as young people age, the osteoclasts in the endosteum break down bone on the internal bone surface, around the medullary cavity [88]. The increase in the medullary canal diameter was reported as a risk factor for implant loosening in younger patients after arthroplasty [89].

I therefore look at results from arthroplasty procedures because the bone-implant interface has been investigated intensively and because the medullary canal is used for implantation in both arthroplasty and DSA.

The loosening of the implanted shaft inside the canal has always been a concern in arthroplasty [90]. Loosening occurs mainly as a result of the cyclical application of axial loads and bending moments during locomotion, which eventually destroy the bond between the implant and the bone [91–92].

A drill bores the tube bone to prepare an area in the medullary canal into which a DSA fixture or prosthesis stem fits exactly. The bone remodeling proceeds from the outer walls toward the interior walls of the medullary canal [93]. Such ossification keeps the implant inside the bone canal by developing multiple microlocks [10].

†The thin layer of cells lining the medullary cavity of a bone.

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Different theories, including the genome-based theory [94], have tried to explain the biological mechanism of loosening of the implanted prostheses, but none are satisfactory [95–97]. Various design modifications of the stems and techniques, with or without use of cement, have been introduced and examined, including taper slip stems with a polished surface, fixation by intramedullary nails, and the use of high-pressure saline to inflate the diameter of a cylindrical implant [98]. However, all known approaches depend on the ability of the medullary canal to act as a holding cavity for the stem of the prosthesis.

One important reason exists why even the most promising technologies addressing loosening in arthroplasty may not be translated successfully to osseointegration. Because a hosting bone is shortened after amputation, its ability to retain a DSA implant is less than a full-length bone's ability to retain a prosthetic stem in arthroplasty. Therefore, the depth of insertion in osseointegration is always less than that following arthroplasty. A model of loading/resistance in total joint replacement and DSA is presented elsewhere [99] and illustrated in Figure 5. The model demonstrated that normal loads from the implanted shaft to the bone walls following DSA will always be greater than following total joint replacement.

For patients with a long residuum, devices with a solid core and a porous collar for osseointegration could be a solution. Two of these devices were applied to attach leg prostheses in dogs. One, designed by Dr. Ola Harrisson and his colleagues [100], had a porous collar on an interlocking nail. The implant was secured with screws inserted from the outside of the bone into the implanted nail, which had a porous collar at the level where it penetrated the skin.

A similar system with a porous tantalum sleeve on the titanium core was applied to a dog with bilateral transtibial amputations. The implants were 123 mm in length with a 43 mm-long tapered threaded stem ranging in diameter from 4 to 6 mm, which was implanted to the tibial medullary canal. The tantalum sleeve was 70 to 80 percent porous by volume, with an average pore diameter of 500 mm. A stem was press-fitted to the tibia with the sleeve providing a stop. At 26 months after initial surgery and 17 months after revision of one of the two implants, the dog's function was restored at a walk, trot, and run [85].

The anchoring and locking approach requires additional operation time and techniques for exact positioning of the screws relative to the holes of the shaft implanted into the medullary canal. Another potential disadvantage of the technique is a predetermined orientation of the porous collar relative to the skin of the residuum, which replicates that seen in the device with the perforated flange [79]. On the other hand, it is a one-step procedure compared with the classical two-step Brånemark procedure, which could be beneficial for patients in the future.

An important argument in favor of the classical two-step procedure is that the after fixture is initially implantated, it remains under the skin of the residuum and is totally "immersed" to the bone. The bone walls keep the fixture unloaded for about 3 months before the second step (installation of the abutment), which is essential for osseointegration. Eliminating the mobility of an implant relative to the bone for the period of bone remodeling is imperative for successful osseointegration. A study by Pilliar and colleagues demonstrated that bone ingrowth can occur in the presence of movement up to 28 μm, while excess movement (150 μm or more) can result in attachment by mature fibrous connective tissue ingrowth [101].

Beyond Brånemark's threaded titanium fixtures for dental implantology (Figure 2), different devices with holes for bone ingrowth to achieve a firm device-bone composite structure have been developed and implemented. In 1913, Dr. Edward J. Greenfield designed implants with cylindrical wire baskets of iridio-platinum. The bone could grow into the

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implant basket body, and it would be secured [102]. A similar approach can be found in Ashukian's Artificial Tooth design [103]. In 1963, Dr. Leonard L. Linkow introduced the "VentPlant" with an open cagelike design that went into the bone with a few threads on a solid body at the top. He then developed a blade implant [104–105]. Linkow's device has a long thin blade that would be surgically placed into a groove in the bone (Figure 6). The long hollowed base of Linkow's implant allows ingrowth of bone, which adds to the device-bone integration [106–107]. Among the negative aspects of the long base is that it requires a lot of space even for a single tooth prosthesis. If the implant has to be removed, a very big portion of bone has to be taken out, which complicates or makes impossible new implantation [108].

No reports, to my knowledge, have tested hollowed systems like Linkow's for DSA of limb prostheses. However, for implanting into tubular bone, the bladelike design can be revisited. I suggested [99] that using the medullary canal contradicts the biological purpose of the canal, namely its role as a designated functional cavity for bone marrow [109]. I also noted that inserting a stem into the canal destroys the endosteum, a thin layer of connective tissue filled with cortical capillaries that lines the medullary cavity. I further developed a hypothesis, to be experimentally verified, that an implant with specially added side elements called "fins" could initiate bone regeneration in the circular direction [110]. I believe that regeneration in the circular direction is biologically more efficient than regeneration in the direction toward the center of the bone canal, which is exploited for cylindrical implants [99]. As a result, the pylon with fins for DSA was recently introduced by Poly-Orth International [110]. The design (Figure 7) remotely replicates Linkow's blade implants [111] and the implants for total hip joint replacement with Wagner's longitudinal ribs [112].

OTHER APPROACHES

The work by Brånemark and his group inspired many animal studies that used the skeleton as a site for modeling different attachment technologies [43,113–114]. As an alternative to titanium, vitallium (alloy of chromium, cobalt, and molybdenum) has been tried extensively since the 1970s [115]. After medullary implantation, porous wafers made of vitallium and carbon showed more fibrous connective tissue than bone ingrowth at the bone-implant interface [116]. With hydroxyapatite (HA)-coating, vitallium demonstrated better bone regeneration and osseointegration than pure titanium [117]. Implants coated with a bone-graft substitute derived from reef-building sea coral coralline replamineform HA were placed in canine subcutaneous tissues to clarify whether the HA matrix acted as a passive matrix for osseous ingrowth when placed in some inherent bone-induction capacity. The implants were well tolerated and elicited no deleterious host response. Connective tissue rapidly infiltrated the pores, but no evidence of bone formation was noted in any of the specimens. It was concluded that this implant material does not act to induce bone formation [118]. Another material tested for percutaneous implantation was nongraphitizing (vitreous) carbon, which has high strength, hardness, and low porosity and permeability [119]. Implants of vitreous carbon with pore diameters 200 to 500 μm were surgically placed in rabbits and pigs [120]. *Staphylococcus aureus* and *Escherichia coli* were inoculated in a test area adjacent to the implant and a remote control area. Results showed that despite a temporarily high rate of colonization and obvious binding of the bacteria to the carbon, the skin-implant interface resists infection by both normal and pathogenic flora.

Mismatch in moduli of elasticity in the bone and the implant is one reason for the loosening and further failure of the implant. The least mismatch so far was achieved by the use of different polymers. Reasonable biointegration and cellular ingrowth were reported, with highporosity expanded polytetrafluoroethylene for facial plastic surgery [121]. Positive

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results were reported with the use of porous polyethylene (Medpor) implants for nasal reconstruction [122].

Many attempts were made to improve or speed up creation of a cell-implant bond by HA coating. HA is a form of calcium phosphate closely related to the mineral phase of bone and teeth, having a similar crystalline structure [123]. Attachment of the bone to the coated titanium implants occurs faster than to the uncoated implants [124]. Mechanical properties and adhesion of the cells to the layer of coating depend on a calcium phosphoric ratio in HA and on the method of coating. The plasma-spraying technique is commonly used to coat the orthopedic implants with HA, but it provides a low bond strength compared with the electron beam deposition method [125]. Different coating techniques result in different bond strengths between the implant and the bone. Coating degrades with time, leading to detachment and subsequent loosening of the prosthesis [126].

While highly biocompatible, HA-coated titanium implants more easily developed a more severe infection in the presence of bacteria than non-HA-coated titanium implants [127].

To improve interface with the cells, titanium implants have been treated with an enamel matrix derivative (EMD) and a vehicle gel propylene glycol alginate [128]. All osteoblast activity indicators were significantly greater in the porous coated region than in the host bone region [129]. The results obtained showed that the EMD treatment did not benefit the bone formation around titanium implants.

Remodeling of extracellular matrix-associated tissue promotes angiogenesis, recruitment of circulating progenitor cells, rapid scaffold degradation and constructive remodeling of damaged tissues, and adhesion and migration of cells [130–131]. Since treating the surface of the implant by the extracellular matrix components has the potential for better cell growth [132], it may be applied in the future for the implantable prosthetic pylon as well. An in vitro study with titanium foam for spine fusion showed the ingrowth of human osteoblasts to the metal structure [53]. To prevent bacteria from colonizing implants and forming biofilms, Antoci and colleagues have covalently attached antibiotics to implant surfaces. They reported that vancomycin-modified implants showed better inhibition of bacterial attachment and proliferation than control titanium surfaces [133].

Porous circumferential components used in total joint replacement, which are advantageous in the bone-device interface [129,134], can be a source of infection for the skin-device zone [135]. Efforts to optimize the roughness of the abutment to achieve a reliable and infection-free skin-device interface were unsuccessful [42], and current clinical studies use abutments with a smooth surface [86].

DISCUSSION

At present, no follow-up data on skin seals around porous percutaneous implants have been collected for more than several months. Reports of successful DSA in humans refer to implanted abutments with a smooth surface [10–86], data on which have been collected for many years. Potential skin mobility is addressed surgically during the second step of the two-step DSA procedure: the removal of the fat layer induces skin ingrowth into the open end of the residuum bone. Patients then get instructions for daily hygienic care of the area where the abutment goes through the skin. Nevertheless, this approach should not be considered the ultimate solution, since it does not reliably protect against skin infection and sinus tract infection affecting bone and other subcutaneous tissues.

Deep ingrowth of skin through the porous pylon may create a safer and more sustainable seal. A potential mechanical advantage exists as well. The total integration of the skin with a

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porous implant reduces skin mobility and therefore reduces the probability of avulsion. Moreover, once the implant is invaded with host tissue, the bacteria are less apt to establish infection [136].

At least two immediate issues exist with total skin integration. The first is the mismatch between the moduli of skin and the implant. The second is the dynamics of initial ingrowth: before its completion bacteria can invade the host tissue [136].

The mismatch issue requires a mediator or mediators between the skin and the implant to be found, with the modulus or moduli gradually increasing within the skinimplant gap [38,137]. Ideally, the mediator layer would function as the eponychium (cuticle) and hyponychium that comprise the junction between the skin stratum corneum and the base of the nail plate [138]. To facilitate the initial ingrowth, various growth factors can be considered [139–142].

The bone-implant interface is another area requiring more study. The current DSA technology, introduced by Brånemark, and arthroplasty use the medullary canal of residual bone as the hosting cavity for implantation. The principal difference between the two procedures lies in the prosthesis design. The prosthesis used for total joint replacement is asymmetric relative to its longitudinal axis, since it has a portion that models the bone head. Two parameters must be controlled for precise installation: axial translation into the canal and angular position of the head of the artificial bone. For that reason, the placement of the shaft of the prosthesis is a press-fit type. In contrast, DSA implants are symmetrical relative to the longitudinal axis and only the depth of installation needs to be controlled, as with dental implants. Therefore, the implants for DSA can be screwed into the canal, a technique that has proven successful in dental implantology.

A screw is a more reliable fastener than a nail. So, too, the screw-in device-bone connection is more mechanically effective than the press-fit type. The screw-in type requires a smaller zone of contact with the hosting bone than the press-fit type for the same value of resistance to detachment. Such a consideration is especially important for DSA in the case of a short residuum.

To summarize the origins of current DSA technology [143]: DSA uses the screw-in method adopted from dental implantology [108] and uses the medullary canal for implantation like arthroplasty [144].

The bond between the bone and all existing implants relies on ossification inside the canal in the inward direction, which is the least efficient when compared with ossification in the longitudinal and circular directions [99]. Inward ossification is induced by damage to the inner walls of the canal during installation of the implant but is limited by the space designated for bone marrow. Longitudinal ossification, introduced by Ilizarov, can lengthen the bone up to 30 percent via distractional osteogenesis [145]. Circular ossification is a mechanism for fracture healing and can widen the bone up to more than two initial diameters [145]. A hypothesis was developed that circular ossification could be used for a more reliable fixation of the implants for DSA [99], and a design of such an implant was introduced [110]. If confirmed experimentally, this approach may decrease loosening both in DSA and arthroplasty. Specifically, circular ossification will counter the increase in diameter of the medullary canal, which diminishes the long-term reliability of the bond between the inner walls of the canal and the implant. The increase in diameter is a natural process in bone development and can be accelerated by a high level of physical activity [146].

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CONCLUSIONS

The increased number of young active persons with amputation due to recent wars has brought more attention to the technology of DSA of limb prostheses. A multidisciplinary revisiting of all aspects of the technology responsible for lifelong safe functioning is now required. For DSA implants, titanium and tantalum remain the most promising materials. They maximize cell attachment, both at the interface between the implant and skin at the transcutaneous passage and between the implant and the residuum bone. Porous implants have a strong potential for improved skin and bone integration if problems of strength and material mismatch can be resolved.

Abbreviations

DSA	direct skeletal attachment
EMD	enamel matrix derivative
HA	hydroxyapatite
TiO_2	titanium oxide

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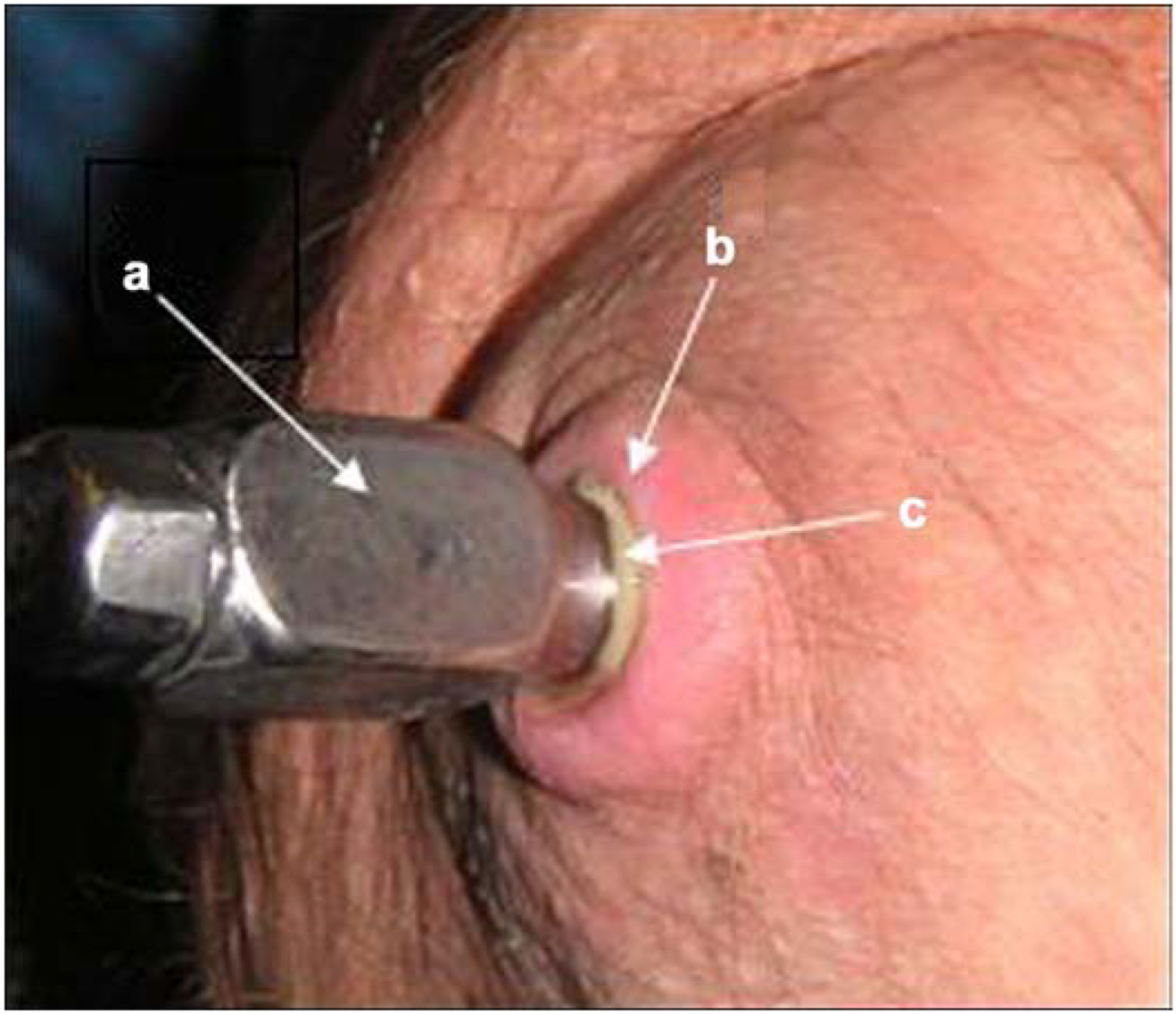

Figure 1.
(a) Titanium abutment penetrating residuum skin, **(b)** surrounding skin, and **(c)** layer of pus between skin and abutment. *Source*: Reprinted from Pitkin M, Raykhtsaum G, Galibin OV, Protasov MV, Chihovskaya J V, Belyaeva I G. Skin and bone integrated prosthetic pylon: A pilot animal study. J Rehabil Res Dev. 2006;43(4):573–80. [PMID: 17123195] DOI: 10.1682/JRRD.2005.05.0160

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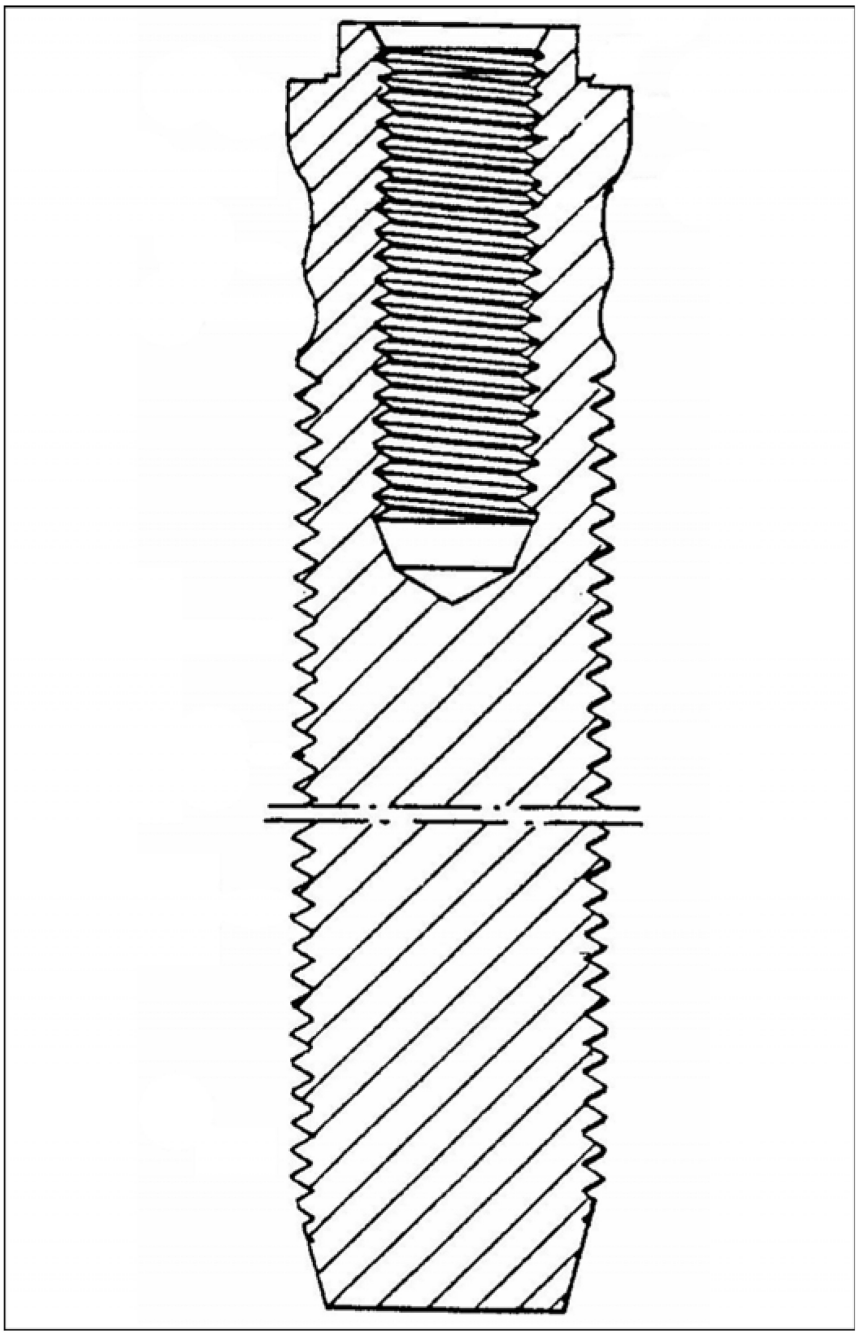

Figure 2.
Implant fixture for tooth prosthesis by Brånemark. *Source*: Adapted from Brånemark P. Implant fixture for tooth prosthesis. United States patent US 4988299. 1991 Jan 29.

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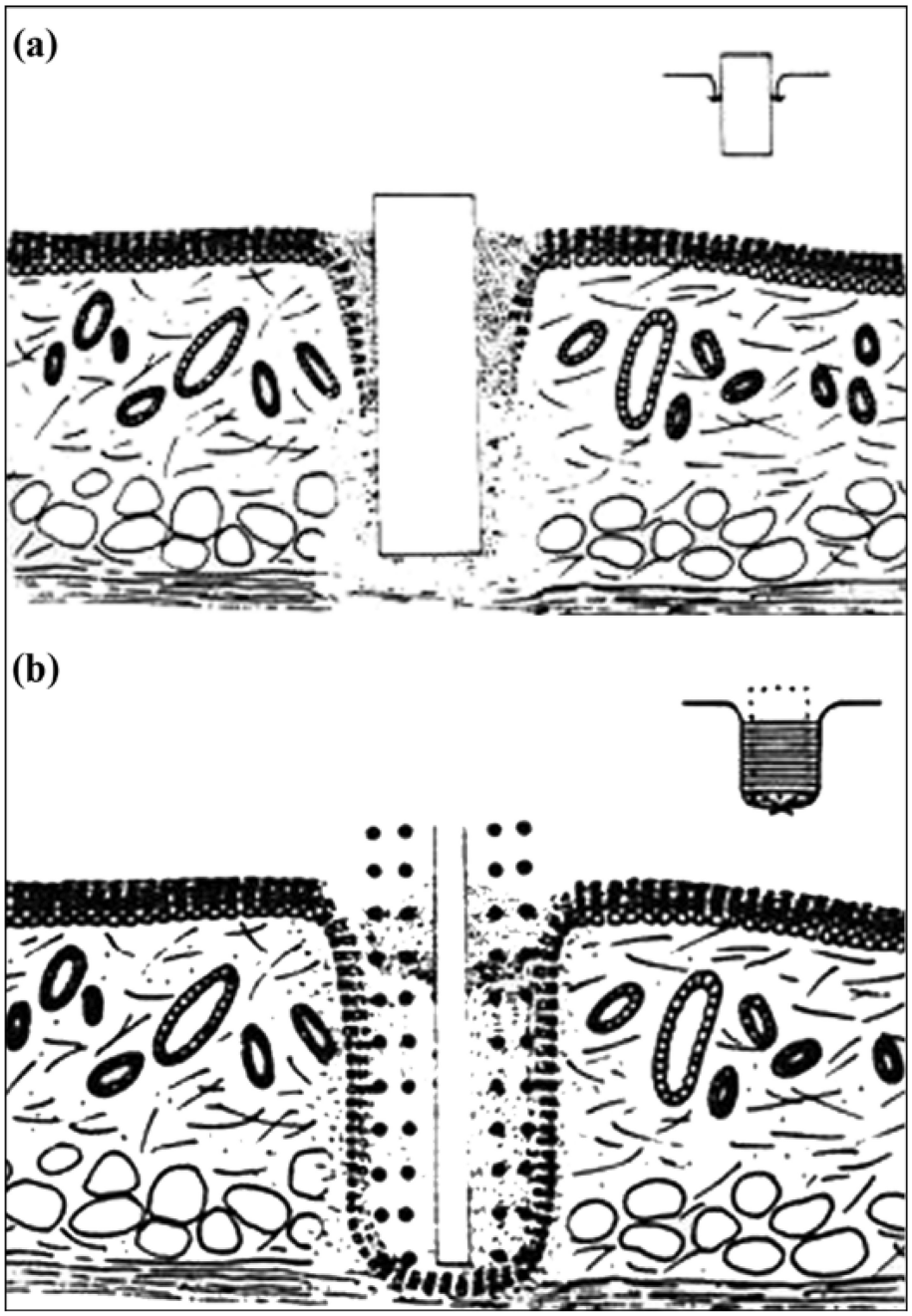

Figure 3.
(a) Schematic of histological section of skin with smooth surface implant. Epidermis migrates and proliferates inward along implant causing marsupialization. **(b)** Epidermal migration through porosity or permigration. *Source*: Adapted by permission from Von Recum AF. Applications and failure modes of percutaneous devices: A review. J Biomed Mater Res. 1984;18(4):323–36. [PMID: 6234317] DOI:10.1002/jbm.820180403

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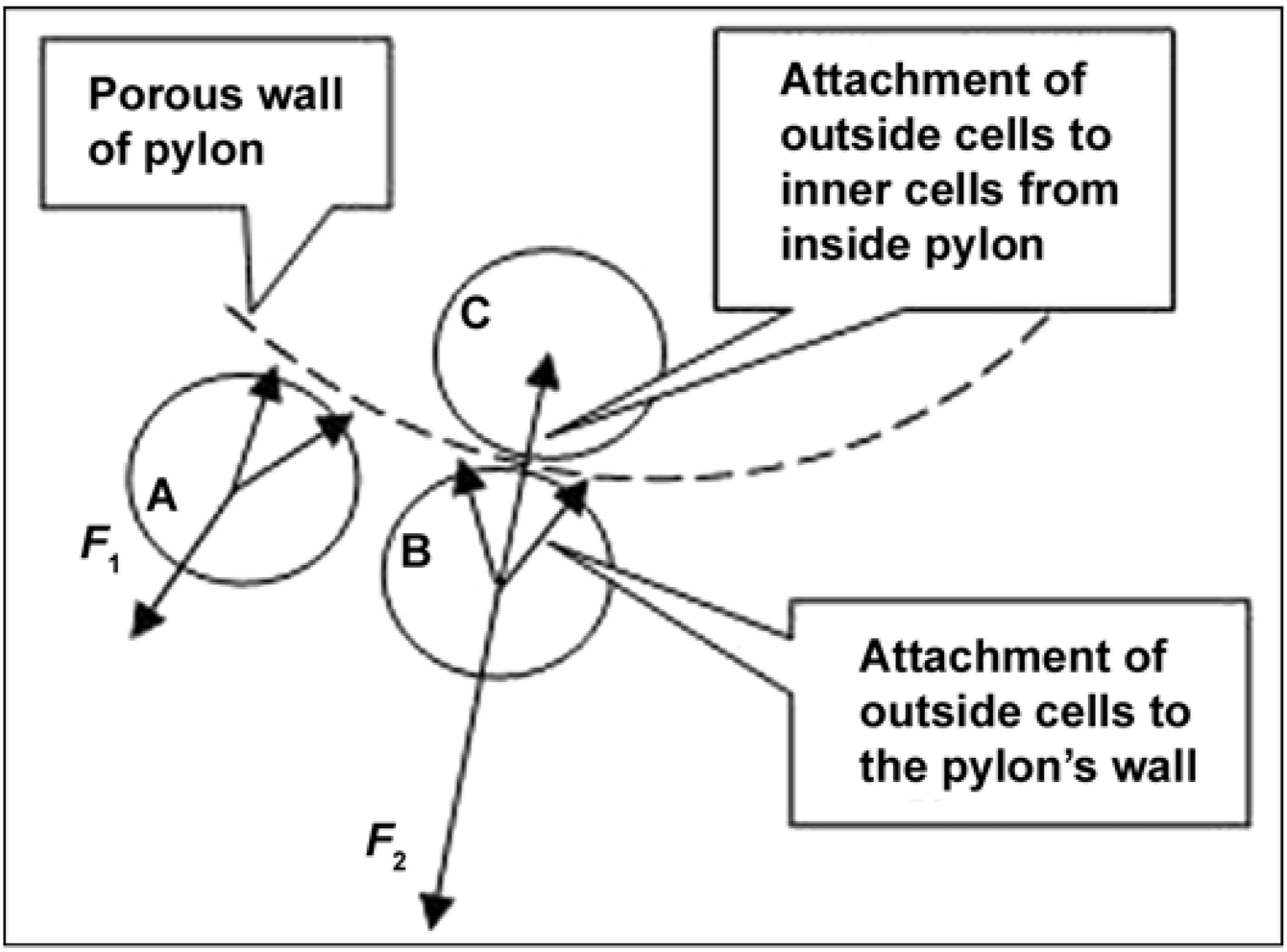

Figure 4.
Minimal force needed for detachment of skin cells A, B, and C from pylon. F_1 = "cell-to-wall" detachment (prior art), F_2 = "cell-to-wall-to-inner cell" detachment in skin and bone integrated pylon design. *Source*: Adapted from Pitkin M, Raykhtsaum G, Pilling J, Galibin OV, Protasov MV, Chihovskaya JV, Belyaeva IG, Blinova MI, Yudintseva NM, Potokin IL, Pinaev GP, Moxson V, Duz V. Porous composite prosthetic pylon for integration with skin and bone. J Rehabil Res Dev. 2007;44(5):723–38. [PMID: 17943684] DOI:10.1682/JRRD.2006.12.0160

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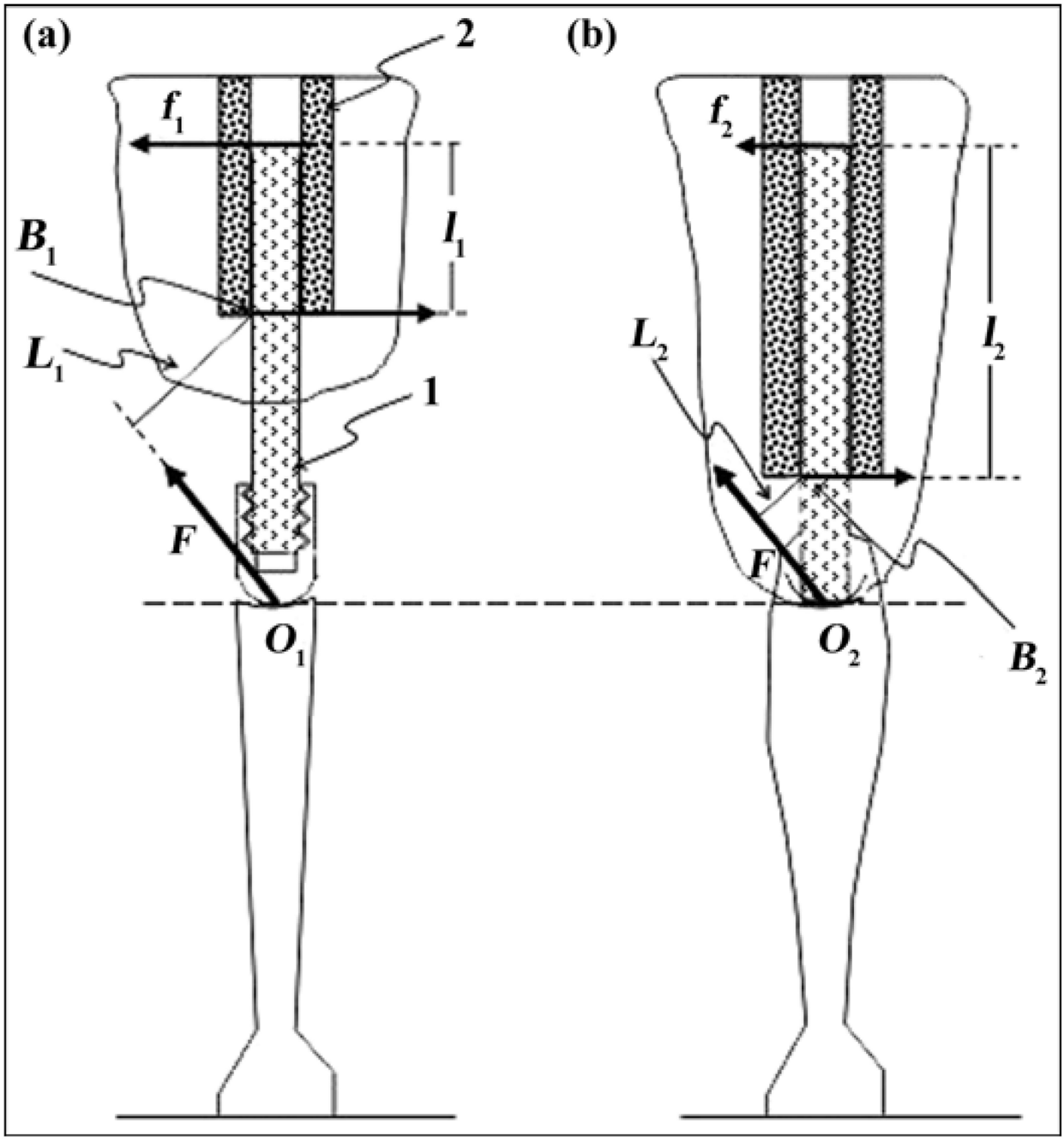

Figure 5.
Modeling of moment of resistance to bending of pylon (1) implanted to medullary canal (2) via **(a)** osseointegration and **(b)** arthroplasty. L_1 = lever arm with respect to point B_1 of bending force F applied to point O_1, L_2 = lever arm with respect to point B_2 of force F applied to point O_2, l_1 = lever arm of reaction force f_1 with respect to point B_1, l_2 = lever arm of reaction force f_2 with respect to point B_2. *Source*: Reprinted from Pitkin M. One lesson from arthroplasty to osseointe-gration in search for better fixation of in-bone implanted prosthesis. J Rehabil Res Dev. 2008;45(4):vii–xiv. [PMID: 18712634]

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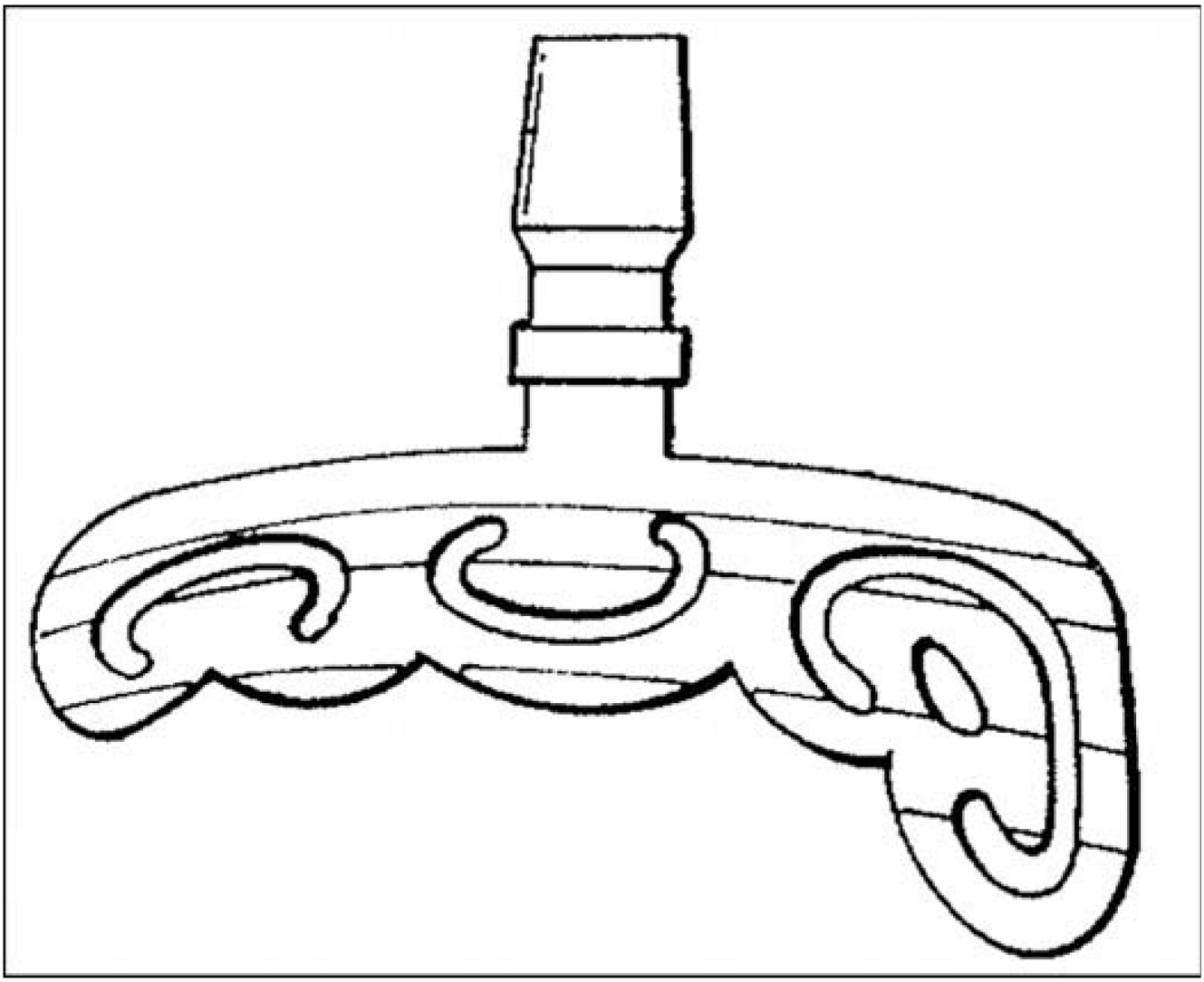

Figure 6.
Linkow's bone adapting tissue packing post system. *Source*: Adapted from Linkow LI, inventor; Oratronics, Inc, assignee. Bone adapting tissue packing post system. United States patent US 3849888. 1974 Nov 26.

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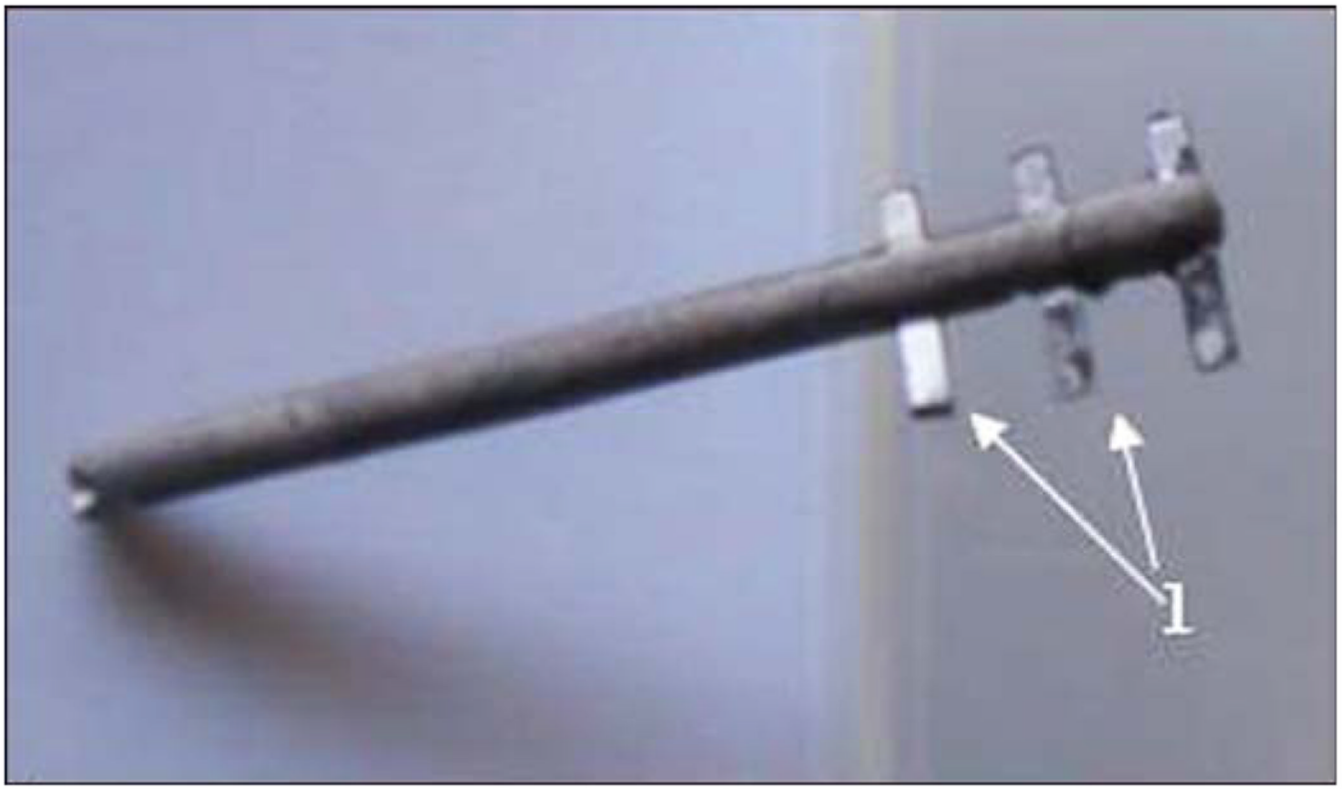

Figure 7.
Poly-Orth International pylon with three pairs of fins (1).