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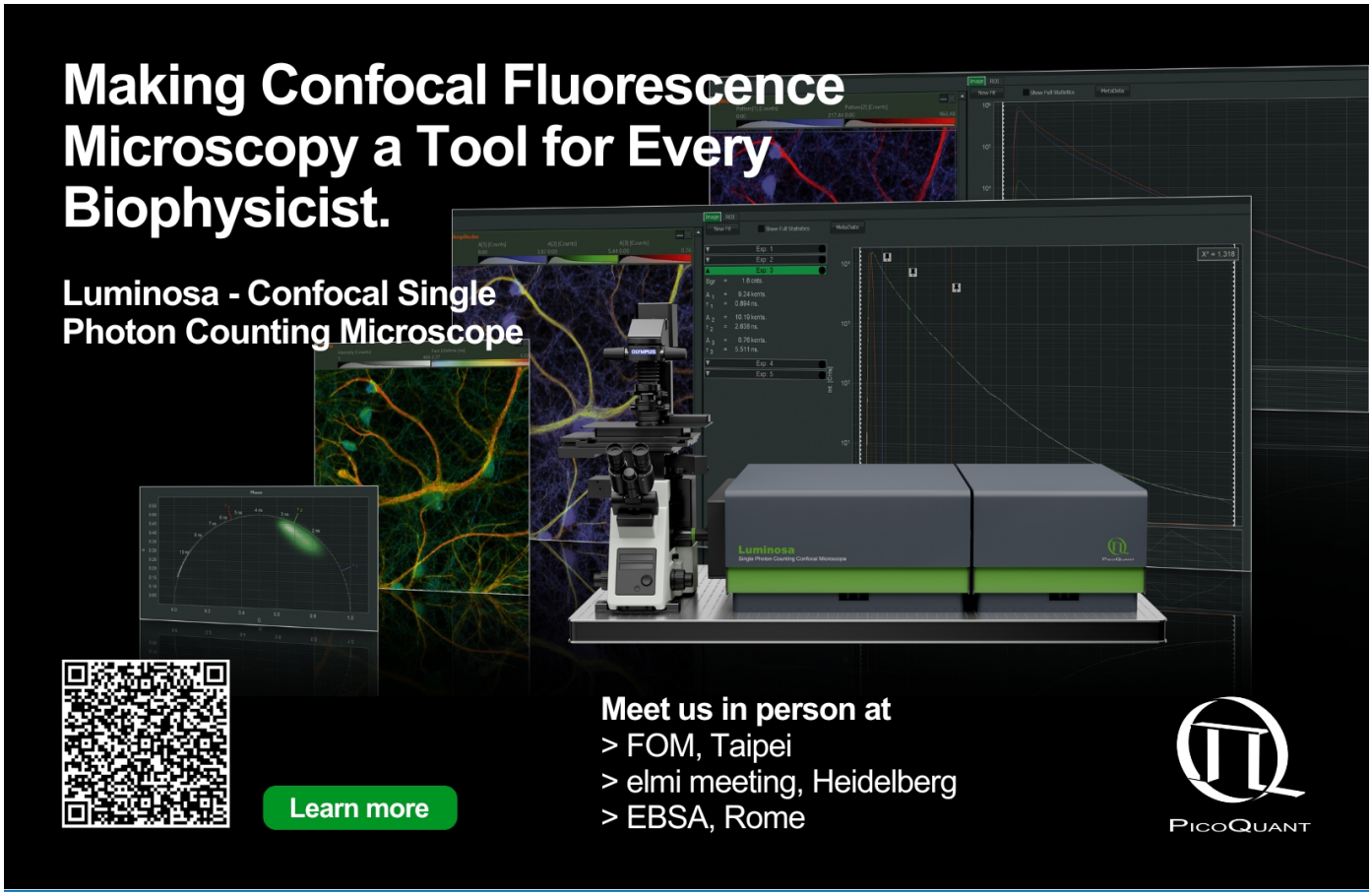

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TOPICAL REVIEW

Progress of surface modifications of temporary anchorage devices: a review

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Keywords: surface modification, temporary anchorage devices, stability, success rates

Abstract

Temporary anchorage devices (TADs) have evolved as useful anchorage providers for orthodontic tooth movements. To improve the stability of TADs, a number of modifications on their surface have been developed and investigated. This review comprehensively summarizes recent findings of clinically applied surface modifications of TADs and compared the biological improvement of these modifications. We focused on sandblasting, large-grit, acid etching (SLA), anodic oxidation (AO) and ultraviolet photofunctionalization (UVP). *In vitro, in vivo* and clinical studies of these surface modifications on TADs with clear explanations, low possibility of bias and published in English were included. Studies demonstrated that SLA, AO and UVP enhance cell attachment, proliferation, and differentiation *in vitro*. The biocompatibility and osteoconductivity of TAD surface are improved *in vivo*. However, in clinical studies, the changes are generally not so impressive. Furthermore, this review highlights the promising potential in combinations of different modifications. In addition, some other surface modifications, for instance, the biomimetic calcium phosphate coating, deserve to be proposed as future strategies.

Abbreviation list

Abbreviation	Definition
AO	Anodic oxidation
BIC	Bone to implant contact
BioCaP	Biomimetic calcium phosphate
BMP2	Bone morphogenetic protein 2
MS	Machined surface
PTV	Periotest values
Ra	Average roughness
ROS	Reactive oxygen species
RTV	Removal torque values/ reverse torque values
SLA	Sandblasting (large-grit), acid etching
ModSLA	Chemically modified SLA surface
TAD	Temporary anchorage device
Ti	Titanium
Ti_6Al_4V	Titanium-6 aluminum-4 vanadium
TiO_2	Titanium dioxide
UV	Ultraviolet
UVP	Ultraviolet photofunctionalization
WCA	Water contact angle

1. Introduction

TADs are widely used to provide stationary anchorage for orthodontic tooth movement. Most TADs are

made from titanium alloys (titanium-6 aluminum-4 vanadium, Ti_6Al_4V) with a conventional MS [1, 2]. The term 'machined surface' is often used as a description of a turned, milled or sometimes a polished surface [3], with an average roughness (Ra) of around 0.08 ± 0.04 μm [4]. The overall stability (red) of TADs is composed of primary stability (blue) and secondary stability (green) [5, 6], as shown in figure 1(A). Primary stability is defined as the interlocking between bone and screw thread, and influenced by alveolar bone density, surface characteristics, together with length, shape, diameter, and rim properties of TADs [7]. It decreases faster than the development of the secondary stability [8, 9]. This leads to a temporary decrease in total stability at the fourth week after TAD placement [9]. Therefore, some researchers suggest to start loading TAD's after week 4, when the secondary stability, which usually refers to the establishment of a direct connection between the TAD surface and living bone tissue [10–13], becomes evident due to new bone formation, and thereafter plays an important role in overall stability [6]. Consequently, a diversity of surface modifications has been applied on TADs to provide an optimum biologic environment for direct attachment of bone tissue to TAD surface (osseointegration) [14, 15], and thus improving secondary stability.

This review focus on clinically available surface modification techniques on TADs, including sandblasting (large-grit) and SLA, AO and UVP and is organized by comprehensively discussion on the modification processes, the biology improvement *in vitro* and *in vivo*, and their clinical outcomes.

To study variables influencing cell adhesion, proliferation and differentiation on the TAD surface, various parameters such as roughness, topography, WCA and chemical composition were addressed [16].

(1) Average surface roughness (R_a): the arithmetic average of absolute values of surface heights and depths (h) relative to the mean base line (figure 1(b)): formula (1)

$$R_a = \frac{\sum h}{l} = \frac{h_1 + h_2 + \ldots + h_n}{n}. \quad (1)$$

R_a of TAD is usually classified as smooth ($R_a < 0.5$ μm), this is the mean for the MS, minimal roughness (R_a 0.5–1.0 μm), moderate roughness (R_a 1.0–2.0 μm) and severe roughness ($R_a > 2.0$ μm) [16]. Moderate rough surfaces (R_a 1.0–2.0 μm) are supposed to encourage the entrapment of fibrin protein and adhesion of osteogenic cells [17].

(2) Surface topography: generally presents three dimensional features that are described as micro–nano-pores, nanotubes, gratings, columns, pits, randomly distributed grooves and so on (figures 2(A)–(D)) [18] all of which may contribute to roughness and wettability of the surfaces of TADs [19]. Features of TAD surface achieved by SLA have been described as large dips, sharp edges, sponge-like structures and small micro pits (figure 2(F)). Features of TAD surface achieved by AO showed uniform micro/nanoscale pore or tube arrays (figures 3(B), 4(G) and (h)) [20]. Compared with smooth surfaces, complex topographical features provide more surface area for interaction with the proteins and the surrounding physiological environment and for cell adhesion, cell stimulation and extracellular matrix formation [19].

(3) WCA: the angle formed between the surface and the line tangent to the edge of the waterdrop. A surface is hydrophilic if the WCA is less than 90 degrees (figures 2(C) and (D)) and hydrophobic in case the WCA is more than 90 degrees. Hydrophilic surfaces show high surface energy and allow fluid to easily wet out. Increased hydrophilicity tends to attach reagents and enzymes, induce osteogenic gene expression and osteoblast differentiation [21].

(4) Surface chemical composition: the arrangement, type, and ratio of atoms in molecules of chemical substances on the surface. This could be changed by various surface modifications, for instance, the presence of phosphorus in phosphoric acid could promote new bone formation [22].

The following parameters are used to evaluate the effects of surface modifications on TADs' stability, some parameters are involved in finite element analysis, animal studies, as well as in clinical studies.

(1) BIC, osseointegration on the bone–implant interface in histomorphometric analysis, is regarded as a predictor for implant stability and survival [13, 25–29]. It is expressed as a percentage of the bone over the entire surface of the implant divided by the whole implant surface [30]. In clinical successful implants, the process of bone remodeling reached an equilibrium state of BIC% remaining at 58%–60% [31].

(2) Removal torque values/ RTV, the forces required to remove TADs. It is a biomechanical measure, as well as an indirect scale of osseointegration [32–34]. A torque of 4 N cm suggests sufficient anchorage capability for TADs [35].

(3) PTV, the amount of tooth/TADs mobility. The Periotest device is originally designed to provide objective measurement of tooth mobility by assessing damping characteristics of periodontium [36], and is also used to determine TAD stability [37, 38]. PTVs ranged between 4 and 8 for TADs indicates that they are ready for loading [39].

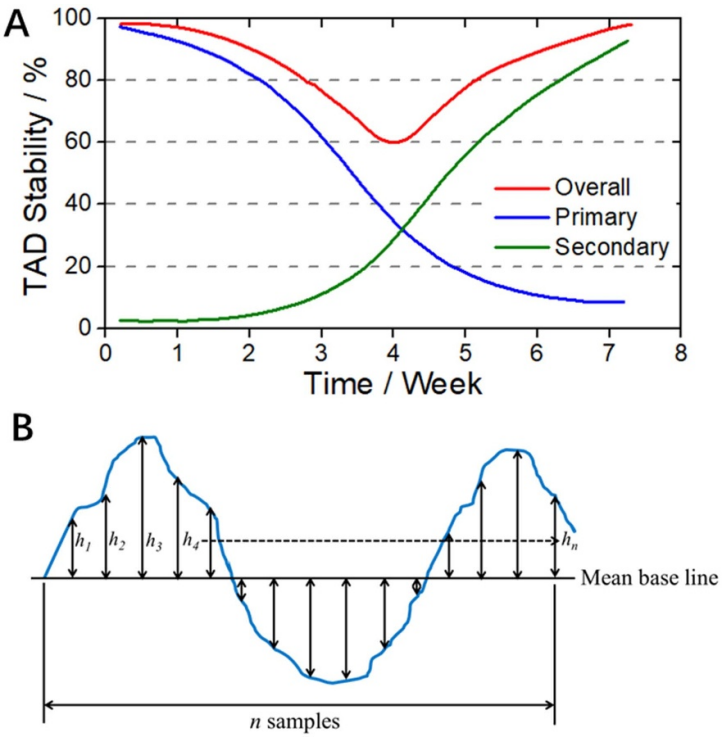

Figure 1. (A) The decrease of primary stability (blue), the increase of secondary stability (green) and the changes of overall stability (red) of TADs through time. Reproduced from [5], with permission from Springer Nature. (B) Definition of average surface roughness (h: absolute values of surface heights or depths relative to the mean base line, l: total sample length, n: number of evenly spaced positions.

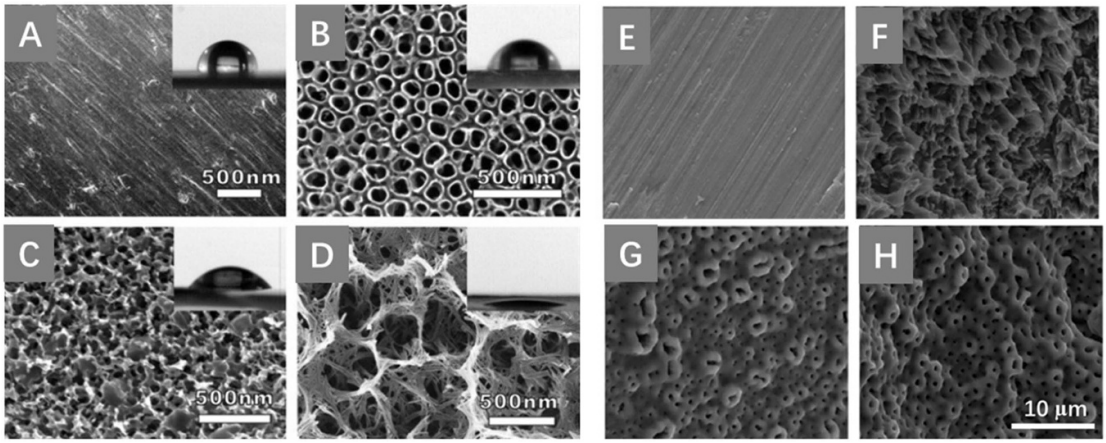

Figure 2. Surface topography of (A) untreated Ti surface, (B) nanotube structure, (C) sponge-like structure and (D) nano/micro nest-like structure, and corresponding contact angle (inset). Copyright (2014), with permission from Elsevier. (A)–(D) Reprinted from [18], Copyright (2013), with permission from Elsevier. Electromicrographs of (E) machined surface (MS), (F) SLA (sandblasted large grits and acid etched resulting in a sponge-like surface with micro pits), (G) AO (anodic oxidation resulting in micropores) and (H) SLA-AO. (E)–(H) Reproduced from [4]. CC BY 4.0.

To compare the improvement in stability by clinically available techniques for TAD surface modification, the above-mentioned parameters based on previous studies were reported and compared in this review article. After which, clinical applicability was addressed and discussed.

2. SLA

2.1. Process of SLA

SLA is a two-step surface modification method (figure 3), whereby the first step is to obtain a primary rough structure of 10–100 μm by sandblasting, using

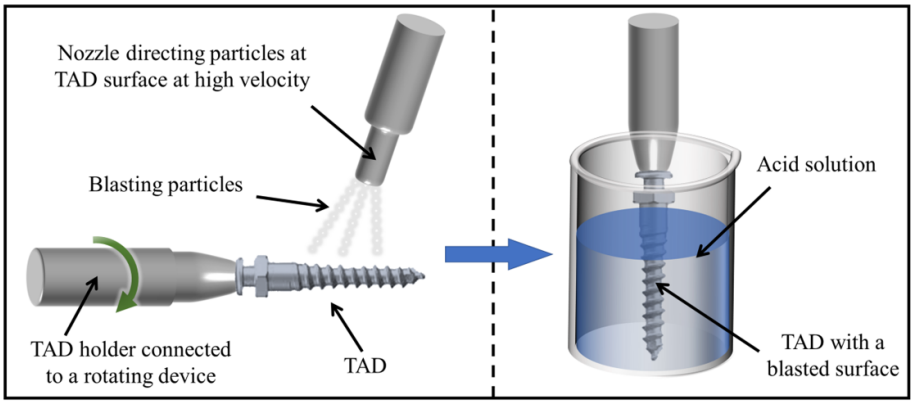

Figure 3. Schematic illustration of the SLA process, including obtaining a primary rough structure by a high-speed jet beam by spraying large-grit particles (left), and a secondary rough structure by acid etching the surface using acids (right).

a high-speed jet beam formed by compressed air to spray different particles (quartz sand, hydroxyapatite, alumina, TiO_2, and iron oxide particles) in a size range between 250 μm and 500 μm (large-grit) onto the TAD surface [40, 41]. The second step is to obtain a secondary rough structure by acid etching the surface at high temperature using acids such as sulfuric, hydrochloric, hydrofluoric and nitric acid, or a mixture of these [42].

The SLA surface achieves a moderate or high Ra of 1–3 μm with large dips, sharp edges and small micro pits (figure 2(f)) [43, 44]. The roughness and topographical features of the TAD surface are considered important factors for the biological response on the bone-implant interface [42].

It is reported that the presence of titanium hydride (TiH_x, $x \leqslant 2$) [45] and the increased oxygen concentrations [46, 47] resulting from a combination of strong acids at high concentrations, endow the SLA surface to be chemically more suitable for binding biomolecules [48].

Furthermore, previous studies indicate that the fluoride incorporation from hydrofluoric acid [48] and the phosphor incorporation from phosphoric acid [22, 47, 49] on SLA surface could promote new bone formation [22, 48].

SLA surface could be further modified by rinsing under nitrogen (N_2) protection and continuously stored in an isotonic sodium chloride (NaCl) solution. This chemically modified clean and active surface, called ModSLA, with increased oxygen and reduced carbon concentration is reported to positively influence the biologic process of TAD osseointegration [50, 51].

2.2. Biology of SLA

The moderate or highly rough SLA surface could improve cell anchorage and provide better osseointegration *in vitro*. According to scanning electron microscopic images, the osteoblast-like cells (MG63) presented a very flat morphology, with few small cytoskeletal processes or pseudopodia on MS while showing an irregular cellular morphology and many pseudopodia that assisted cell attachment on the SLA surface [17]. It has been demonstrated that, SLA surface with increased surface energy can promote osteoblast differentiation [52].

Furthermore, SLA surfaces have been reported to have a potential to shift macrophages towards tissue-inflammatory M1 macrophages, which may have a negative effect on soft tissue (gingival, mucosal) healing. Therefore, SLA surfaces should be kept within the bone [53] and should not contact gingival connective tissues.

The complex topography of SLA surface also mediates secretion of angiogenic growth factor by osteoblasts [54] and induction of osteogenic differentiation [55], promoting a higher and faster osseointegration. This usually results in higher BIC for the unloaded or loaded SLA TADs implanted in tibias and femurs of New Zealand rabbits, as well as in the maxilla of Land Race pigs compared with MS TADs [4, 40]. The increased osteoblastic attachment and differentiation, in turn, enhances RTV [56–60]. Maino *et al* however reported only a significant increase of RTV in the SLA group in rabbits, but no significant change of BIC [61]. This inconsistency may be explained by the large variation between animals.

Chemical modification (Mod) by rinsing under nitrogen (N_2) protection and storing continuously in an isotonic sodium chloride (NaCl) solution, did not change the surface topography, however the ModSLA surface showed a significant decrease in WCA (WCA $=$ 138.3 $\pm$ 4.2 in SLA, WCA $=$ 0 in ModSLA) [51]. In the maxilla of minipigs the increased wettability contributed to a significantly higher BIC in ModSLA dental implants compared with SLA dental implants at 2 weeks (49.30% versus 29.42%) and at 4 weeks (81.91% versus 66.57%).

However, at 8 weeks, similar BICs were observed in both groups. These results support the potential of chemical modification of the SLA TADs to accelerate their osseointegration during the initial stages of bone regeneration [51].

Although the specific values of BIC and RTV may vary in different studies because of differences in time points, animal species, insert locations and other parameters, it can be concluded that SLA and ModSLA surface preparation contribute to (secondary) TAD stability.

2.3. Clinical studies of SLA

Since specific characteristics of soft tissues and bone can influence study results, the outcomes of animal studies cannot be simply generalized to humans. Consequently, clinical studies were carried out to further investigate performance of SLA TADs in humans [62, 63]. According to a split-mouth controlled trial with identical sites (between the second premolar and first molar) and force application (250 g, 6 weeks after insertion), RTV was higher for the SLA group than that of MS group ($p < 0.05$). Furthermore, the survival rate was also higher, but not statistically significant (90.3% to 83.9%) [34]. In another study with immediate loading on TADs, the survival rate was also insignificantly higher in SLA TAD group (93.5% to 82.5%) [64]. These findings suggest that SLA gave some unimpressive improvement. Split-mouth studies with larger sample sizes in a clinical setting are warranted in the future.

3. AO

3.1. Process of AO

To be oxidized, the metal is fixed on the anode of the electrolytic cell for a discharge oxidation process. The implant is immersed as the anode in a homogenous electrolyte containing strong acids such as phosphoric acid (H_3PO_4), ammonium fluoride (NH_4F), sulfuric acid (H_2SO_4), hydrogen fluoride (HF), or nitric acid (HNO_3) or inhomogeneous electrolytes such as calcium acetate, calcium glycerophosphate, NaCl: NH_4F and $NaHSO_4$:HF:NaF with the passage of a high current density or voltage [42, 65].

The oxide film (titanium dioxide, TiO_2) arising from the oxidation process, undergoes a repeated alternating process of formation and dissolution, during the oxidation procedure. This leads to the final formation of uniform micro/nanoscale pore or tube arrays [32, 66] (figures 4(A)–(D)). The anodized surface enhances osseointegration because of mechanical interlocking through bone growth in the pores [67]. The nanotubular and open pores allow for potential incorporation and sustained release of drugs around TADs [68].

Different current voltages [69] or durations of application [70], the modality through which current is provided—such as constant [71] or pulse [65] of the anodizing process [65, 69, 71, 72] have interactive effects on physical, chemical and biological properties of the oxide film [65, 72, 73]. For instance, the film with thickness of 600, 800 and 1000 nm showed significantly stronger bone response to implants than that of 17 and 200 nm. This difference in response was explained by porosity, pore size distribution, and crystallinity of the oxide film (figures 4(E)–(G)) [23].

3.2. Biology of AO

AO resulted in an increase in surface roughness and surface area [23, 32]. This encourages human osteoblast-like MG63 cells to attach, spread and proliferate better, typically containing filopodia- and lamellipodia-like extensions compared to those on the MS (figures 4(H) and (I)) [24, 74, 75].

Oxidation in phosphoric acid solution produced titanium oxide layers with the presence of phosphate [76]. Furthermore, osteoblasts secrete and deposit more calcium onto anodized surface [69, 77]. The enrichment of Ca (from calcium acetate and calcium glycerophosphate) and P (from H_3PO_4) in electrolyte solution during oxidation process on the anodized surface layer is supposed to induce better bone cell functions surrounding TADs [65].

In vivo evaluation, AO layers showed a positive effect on the rapid recruitment of mesenchymal cells, the rapid triggering of gene expression crucial for bone remodeling and the transient nature of inflammation [78]. These biological mechanisms contributed to significantly greater RTV of anodized TADs in rabbit tibias than that of MS [79]. Sung–Hwan Choi however reported that AO TADs showed no clinically significant superiority in biomechanical stability in mandibles of beagle dogs compared with only MS TADs [32]. This inconsistency may have resulted from variations during the AO process.

AO and SLA showed no significant difference in osteointegration according to an investigation on a rabbit femoral bone model [80]. However, sandblasted, large-grit and AO TADs were reported to exhibited a higher RTV than SLA and MS ones in an *in vitro* experiment [81]. Moreover, SLA-AO surface presented higher initial cell adhesion and induced greater filopodia than MS, SLA and AO only groups. In a beagle dog experiment the SLA-AO group revealed greater BIC than the MS, SLA or AO group [4]. Apart from stability, SLA-AO can also reduce the damage due to the insertion of TADs, both in the tissue and TADs [82]. These findings suggest that the combination of SLA and AO surface treatment could be a promising surface modification for TADs.

3.3. Clinical studies of AO

Significantly more bone was found inside the threaded area of the oxidized TADs placed in the mandible and maxilla of human jawbone. This indicates that anodic oxidized TADs show a significantly higher bone response than the MS ones [28].

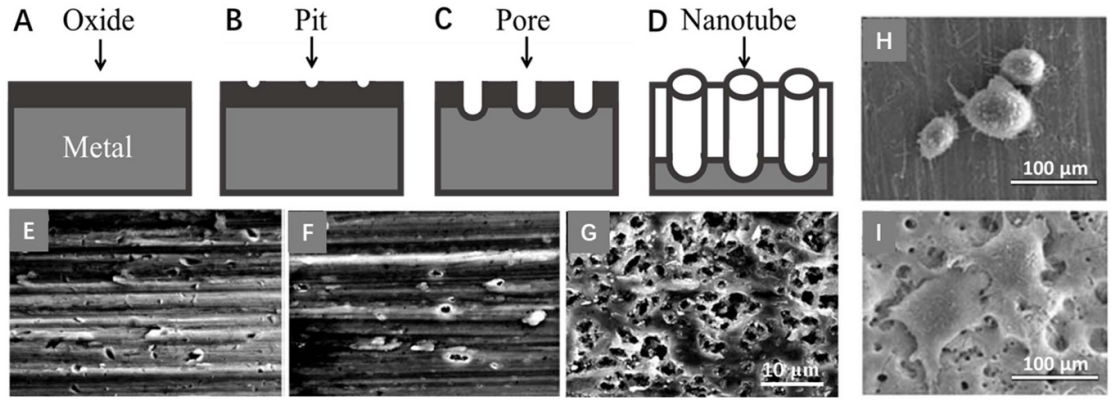

Figure 4. Schematic diagram of anodic oxidation layer formation with prolonged time (A) oxide layer formation, (B) pits formation in the oxide layer, (C) formation of pores in oxide layer, (D) nanotube formation in the oxide layer. Surface morphologies of AO at thickness of (E) 17 nm, (F) 200 nm, and (G) 1000 nm. (E)–(G) [23] John Wiley & Sons. Scanning electron micrographs of MG63 cell on (H) machined and (I) anodized surface. (H), (I) Reproduced from [24], with permission from Springer Nature.

4. UVP

4.1. Process of UVP

TADs can be treated by UV radiation for various time periods up to 48 h to improve their biological properties (15 min is usual). UV radiation is categorized into UVA (wavelength λ = 320–400 nm); UVB (λ = 280–320 nm); and UVC (λ = 200–280 nm) [83, 84]. Generally, UVP of TADs was usually performed by placing them in a custom-fabricated UV chamber consisting of UVA (power: 15 W, wavelength: 3520 nm, intensity: 0.1 mW cm^{-2}) and UVC (power: 15 W, wavelength: 250 $\pm$ 20 nm, intensity: 2.0 mW cm^{-2}) lights for 15 min prior to use [29, 85, 86].

Photocatalytic activity has been regarded as the main mechanism of UVP [27, 87] (figure 5(a)). Photocatalytic activity is achieved by UV photons, which can alter chemical bonds, but cannot ionize atoms and can induce an electronic energy change in molecules [88]. Such change contributes to electrostatic conversion from negative to positive, and the generation of superhydrophilicity on machined, acid-etched and SLA surfaces [89–91]. On titanium dioxide surfaces, the superhydrophilicity is also induced by UV light striking titanium dioxide, exciting an electron which reacts with oxygen in the air to create a superoxide (the major species of ROS), while adsorbed water reacts with the resultant electron hole to create a hydroxide anion, responsible for the superhydrophilic properties [92, 93], as shown in figures 5(B)–(E). Superhydrophilic surfaces are capable of promoting proliferation and differentiation of osteoblasts [94].

The UVP offers two questionable additional advantages. One is the direct hydrocarbon decomposition effect, which could reverse biological aging (the formation of an amorphous, protective oxide coating on Ti surface when exposed to air [95, 96]). However, this effect may be only necessary if TADS are not vacuum packed. Another advantage is the antibacterial property rendered by the accumulation of ROS during UV irradiation process [27, 97]. Because ROS causes partial decomposition of the bacterial cell wall, eventually resulting in cell decomposition and death [98, 99]. However, the clinical effect of ROS may be very unfeasible, as TADs will be invaded by numerous bacteria constantly once used in mouths.

4.2. Biology of UVP

Although the surface morphology showed no obvious change after UVP, the increased surface wettability, electrostatic conversion, antimicrobial effect, and decreased contamination have all been proved to synergistically improve surface biocompatibility and osteoconductivity [94]. UVP was recognized as a trustworthy surface modification without change of surface topography and roughness and could lead to greater biocompatibility of MS.

As shown in the confocal microscopic images in figures 5(B)–(E), osteoblasts were clearly larger with more pseudopodium stretching further in all directions on UV-irradiated surfaces than on untreated ones [27]. UV irradiation not only increased the rates of attachment, but also increased proliferation, osteogenic-factor production, osteogenesis-related gene expression and differentiation of osteoblasts and MC3T3-E1 pre-osteoblasts [94, 97, 100, 101]. On the other hand, UVP was capable of reversing biological ageing of the surface (reducing titanium oxide on the surface). Titanium oxide surface may reduce the attachment of osteoblasts and delay the spread of osteoblasts [100, 102], reversal of biological ageing enables regaining of the surface bioactivity [83, 103]. Consequently, UVP indirectly enhances surface bioactivity.

Furthermore, UV irradiated AO surface may reduce early bacterial attachment, bacterial colony

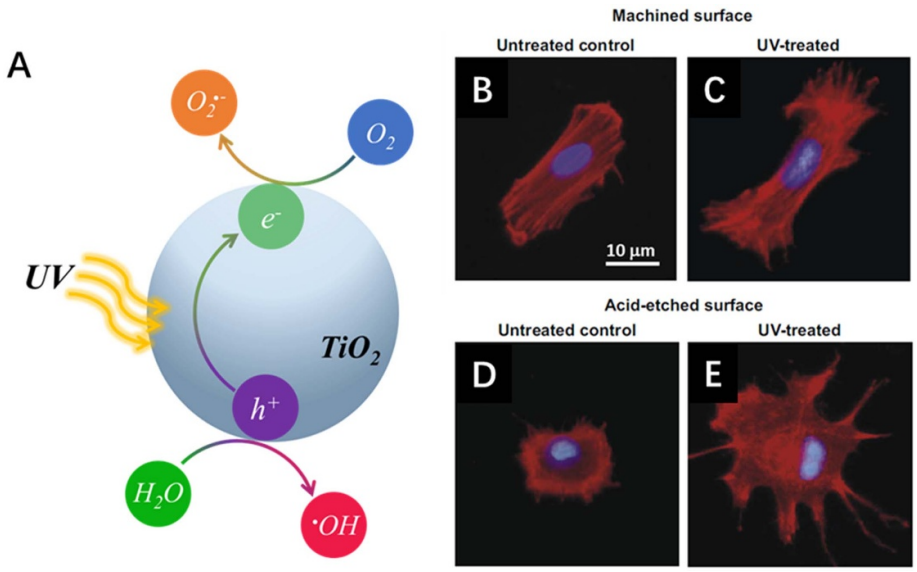

Figure 5. (A) The photocatalytic effect of TiO_2 under UV irradiation. Confocal microscopic images of osteoblasts 3 h after seeding, with dual staining of DAPI for nuclei (blue) and rodamine phalloidin for actin filaments (red), (B) osteoblasts on untreated machined surface, (C) osteoblasts on UV-treated machined surface, (D) osteoblasts on untreated acid-etched surface, (E) osteoblasts on UV-treated acid-etched surface. (B)–(E) Reprinted from [84], Copyright (2009), with permission from Elsevier.

formation, and biofilm formation, without hampering tissue cell integration [104, 105]. Additionally, this surface modification may continue to reduce the bacterial attachment even after implantation and wound closure [87, 104]. As infection is one of the main complications of TADs, UVP is supposed to positively affect success rate of TADs.

In a rat study, up to nearly 100% BIC on the UV irradiated dental implant surface occurred with virtually no intervention by soft tissue 4 weeks after implantation [84]. Such intact BIC was seen as early as 2 weeks in another study, compared with that of control group at 28.6% [106]. This increased osseointegration resulting from UVP may also apply to TADs. In other studies based on TADs, it is reported that UVP significantly increased the RTV and BIC, and decreased PTV [89, 107]. The displacement of untreated screws under a lateral tipping force was greater than that of UVP TADs [86, 89].

4.3. Clinical studies of UVP

UVP was reported to lead to greater biocompatibility of MS, AO or SLA surface *in vitro* and *in vivo* [100]. However, in a split mouth study of 17 patients, there was no significant difference in BIC between UV treated and untreated TADs, revealing that, in this study, UVP did not enhance the biologic potential of titanium TADs in clinical application [29].

5. Discussion

The success rates of TADs range from 57% to 93.43% [108, 109]. These large variations can be explained by the criteria of success, which could be success without mobility, success with mobility but could fulfill all objectives, or success with displacement but could fulfill all objectives [110]. On the other hand, patients with risk factors are prone to suffer TAD failure. These risk factors include: under the age of 20 years old, smoking, poor oral hygiene, short TADs (<8 mm), TADs used for intrusion and several more factors [111–113]. In these cases, it is valuable and necessary to make some improvements on TADs by surface modifications with an aim to adapt TADs to requirements of the orthodontic technique.

Table 1 displays an overview of the effect of these modifications on the surface and on the clinical effectiveness of TAD's. SLA has been used commonly for dental implant surface modification for the past 3–4 decades [114, 115]. This process achieves an optimal roughness, leading to early host tissue response required for faster osseointegration [116], which can serve as an effective primary mechanical modification on MS. AO provides additional advantage to SLA, as a drug storage and delivery system [117], which is suitable for the cases with drug-carrier capabilities, regardless of MS or SLA surface. Moreover, the rate and duration of the drug release, as well as the pore diameters and the thickness of the oxide layer can be controlled by changing anodization conditions [118]. Although UVP does not significantly change surface morphology of TADs, it increases surface wettability, decreases contamination, and has antimicrobial effect on the TAD surface [104, 119]. Therefore, UVP can be regarded as an effective multi-functional surface modification, especially for the personalized cases, parameters and durations can be set in chair side, because only a chamber is needed for UVP.

Table 1. Summary of physical properties, *in vitro, in vivo*, and clinical studies of SLA, AO and UVP on TADs.

Aspect		Surface modification techniques		
		SLA: sandblasting large grit acid etching	AO: anodic oxidation	UVP: ultraviolet irradiation
Physical properties	Average roughness	Moderate or high rough	Several dozen to several hundred nanometers	No change
	Surface topography	Large dips, sharp edges, small micro pits	Micro/ nanoscale pore or tube arrays	No change
	Wettability	Hydrophilic	Hydrophilic	Super hydrophilic
In vitro studies	Cell spreading	Increased	Increased	Increased
	Cell proliferation	Increased	Increased	Increased
	Cell differentiation	Increased	Increased	Increased
In vivo studies	BIC	Increased/ no significant difference	No significant difference	Increased
	RTV	Increased	Increased/ no significant difference	Increased
	PTV	Not mentioned	Not mentioned	Decreased
Clinical studies		Increased RTV Increased survival rate	Increased BIC	No significant difference

SLA, AO and UVP are all well-established technologies in manufacture now [28, 29, 34, 64], and therefore there is no challenge in translational application generally. Among them, the UVP is more feasible, since a chamber consisting of UV light is available chair side in clinical application, and personalized designs for patients can be realized. It can be anticipated that SLA and AO will also be applicable in clinics in the future.

Although SLA, AO and UV have shown to be capable of inducing a remarkable improvement in experimental research, the clinical effectiveness is controversial [28, 29, 34]. On one hand, histologic evaluation is lacking because it was a human study [34]. One the other hand, the individual differences and small sample sizes. Concerning this, sample size calculation should be performed cautiously in research design, taking mainly type I error, type II error, effect size, outcome's variability into consideration [120]. More future clinical studies with larger sample sizes are warranted. Furthermore, a combination of different processes, for instance, SLA and AO or AO and UVP, deserves to be taken into consideration, since combined they showed more significant improvement than that of any single process [4, 80, 81].

There are many other promising surface modification techniques, which have not been applied clinically yet, for instance, the biomimetic calcium phosphate (BioCaP) coating reported by Liu. It is prepared under physical condition (37 °C, pH = 7.4) and the bioactive agents, for instance BMP2 can be incorporated during the preparation of the BioCaP coating. The BioCaP coating has been demonstrated to increase osteoconductivity of TADs, and the incorporated BMP2 can be sustained released with the biodegradation of the BioCaP coating [13, 121–123]. Furthermore, BIC of dental implants was increased by BMP2 incorporated BioCaP coating *in vivo* [124, 125]. Considering that TADs are comparable to dental implants, this BioCaP coating would be a future proposal to improve bioactive properties of TAD surface.

6. Conclusions

SLA, AO and UVP can improve biocompatibility and osteoconductivity of TADs *in vitro* and *in vivo*. More clinical studies are warranted in the future. Searching for the optimal combination of these procedures seems very useful.

Data availability statement

All data that support the findings of this study are included within the article (and any supplementary files).

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Author contribution statement

Menghong Li contributed to concept and design, drafted and critically revised the manuscript; Reinder Kuitert contributed to design, drafted and critically revised the manuscript; Yuanyuan Sun, Mingjie Wang, Wen Kou and Min Hu contributed to data interpretation, critically revised the manuscript; Yuelian Liu contributed to conception and design, critically revised the manuscript. All authors gave final approval and agree to be accountable for all aspects of the work.

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