Engineering Science and Technology, an International Journal 36 (2022) 101277

HOSTED BY

ELSEVIER

Contents lists available at ScienceDirect

Engineering Science and Technology, an International Journal

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

Review

Nickel titanium alloys as orthodontic archwires: A narrative review

Idil Uysal [a], Bengi Yilmaz [b], Aykan Onur Atilla [c], Zafer Evis [a,d,*]

[a] Middle East Technical University, Department of Biomedical Engineering, 06800 Ankara, Turkey
[b] University of Health Sciences Turkey, Department of Biomaterials, 34668 Istanbul, Turkey
[c] Cyprus International University, Faculty of Dentistry, 99258 Nicosia, Cyprus
[d] Middle East Technical University, Department of Engineering Sciences, 06800 Ankara, Turkey

A R T I C L E I N F O

Article history:
Received 23 May 2022
Revised 17 September 2022
Accepted 16 October 2022
Available online 5 November 2022

Keywords:
Nickel titanium archwires
Friction
Corrosion
Superelasticity
Biofilm formation

A B S T R A C T

Nickel-titanium (NiTi) archwires have been widely used in orthodontic treatments at the aligning and leveling step which is the initial stage of the treatment. This review presents the concepts related to NiTi archwires from the materials science and engineering perspective. Analysis of the research profile showed that mechanical properties as a research topic constituted 25.9 % of the research articles published in last two decades. The effects of processing parameters, stress and temperature on phase transformations and corrosion are summarized. The main factors affecting corrosion are stress, temperature and pH of the environment. The effects of surface roughness, brackets and environment on friction are discussed. The surface modification studies are reviewed. The consequences of deflection heat treatment, aging on mechanical properties are described. NiTi archwires are also evaluated in terms of biological properties from a clinical perspective. Clinical studies have focused on mechanical properties and effects on tooth alignment, biofilm formation and Ni ion release. It is anticipated that the studies that will shape the future of the NiTi arcwire field will be on welding of different types of archwires, computational methods, and improvement in coating technologies.

Contents

1. Introduction 2
2. Structural properties 2
   2.1. The effects of processing parameters on phase transformations 3
   2.2. The effects of stress on phase transformations 3
   2.3. The effects of temperature on phase transformations 4
3. Surface properties and modification 4
4. Friction 5
   4.1. The effects of brackets on friction 6
   4.2. The effects of environment on friction. 6
5. Corrosion 6
   5.1. The effects of environment on corrosion 7
   5.2. The effects of stress on corrosion. 7
   5.3. The effects of temperature on corrosion 8
6. Mechanical properties 8
   6.1. The superelastic property 8
   6.2. The effects of deflection on mechanical properties. 8
   6.3. The effects of heat treatment and aging on mechanical properties 9
   6.4. The effects of environment on mechanical properties 9
7. Biological properties. 9

* Corresponding author at: Middle East Technical University, Department of Engineering Sciences, 06800 Ankara, Turkey.
E-mail address: evis@metu.edu.tr (Z. Evis).
Peer review under responsibility of Karabuk University.

https://doi.org/10.1016/j.jestch.2022.101277
2215-0986/© 2022 Karabuk University. Publishing services by Elsevier B.V.

8. Clinical perspectives . . . 10
8.1. Mechanical properties in clinical studies . . . 11
8.2. Biofilm formation . . . 11
8.3. Ni ion release . . . 11
9. Computational studies . . . 11
10. Future perspectives . . . 12
11. Conclusion . . . 12
Declaration of Competing Interest . . . 12
References . . . 12

1. Introduction

Nickel-titanium (NiTi) alloys with different properties, such as shape memory and superelasticity, have been widely used as archwires for orthodontic treatments and as root canal instruments for endodontic treatments due to their ability to apply light and continuous force on malpositioned teeth with a wide range of displacements [1,2]. In addition to archwires, orthodontic tools, such as closed T loops and NiTi coil springs were studied widely [3,4].

NiTi alloys were first used in the 1970 s in orthodontic applications. The first composition consisted of 50 % Ni and 50 % Ti. The stabilized martensitic form was commercialized before the superelastic property of the material was developed. In the mid-1980 s, the superelastic property of the material was investigated by the stress-induced transformation of active austenitic alloy to martensite. Copper and chromium were incorporated into NiTi alloy in the mid-1990s [5]. NiTi alloys used in orthodontic archwires were classified into four groups: martensitic stabilized with reduced stiffness, austenitic-active with superelastic property, martensitic heat-activated, and martensitic force activated [1].

The number of research articles published between 1981 and 2022 (March) on Web of Science was determined as 407 by using the keywords “nickel titanium”, “archwire”, “NiTi”. The percentage of research topics are given in Table 1.

Table 1
Percentage of research articles between the years 1981 and 2022 based on research topics.

Research topic	Percentage of research articles between 1981 and 2022
Mechanical properties	25.9
The degree of tooth alignment	14.9
Friction	13.5
Corrosion	12.8
Surface properties	10.9
Structural properties	6.7
Biological properties	6.1
Pain perception	5.0
Dimensional properties	1.3
Thermal properties	1.1
Physiological properties	1.1
Magnetic properties, Duration, Phonetics, Aestetic perception	0.8

2. Structural properties

The structural properties of NiTi archwires are investigated mainly by the phase transformation studies, which have a strong relationship with mechanical properties. Shape memory and superelasticity, which occur with the transition from austenitic to martensitic phase, are the two prominent properties of NiTi alloys used in orthodontic applications (Fig. 1).

Between the austenite and martensite phases, there is a transition to R-phase according to differential scanning calorimetry (DSC) studies [6]. R-phase was detected in samples with shape memory property such as Neo Sentalloy®, and 35 °C Copper Ni-Ti™ during the reverse transformation from austenitic to martensitic phase [7,8]. Dislocations and precipitates formed R-phase that provided excellent mechanical properties such as fatigue resistance [8]. Unlike in DSC, R-phase peaks were not observed in XRD since austenite and martensite peaks obscure them [6].

The phase transformation from austenite to martensite is observed by heat treatment, applied stress, or both (thermome-

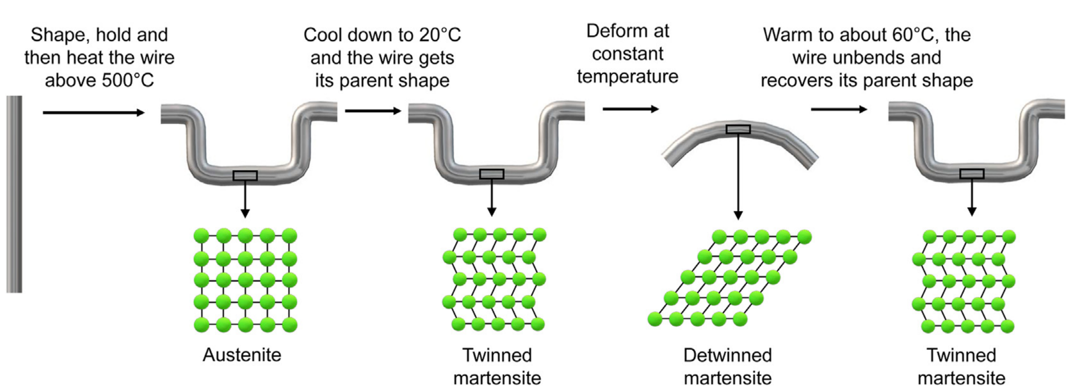

Fig. 1. A simplified overview of the shape memory effect and the phase transitions. A shape memory NiTi wire can be treated at high temperatures to set a predetermined parent shape. When the wire is cooled, the phase changes from austenite to martensite. If the wire is deformed upon loading, the applied stress reaches a critical value for de-twinning. When heat is reapplied, the wire recovers its parent shape during the transition from martensite to austenite phase, and the crystal structure returns to martensite upon cooling down to 20 °C.

chanical treatment). Stable at approximately 100 °C, austenite is the parent phase [9]. As it cooled down, martensitic (daughter phase) transformation occurs within the critical transformation temperature range. Changes in the electron bonding and lattice structure of the NiTi alter its properties, such as stiffness, elastic modulus, and electrical resistivity, as a function of temperature. At 25 °C, austenite peaks (200), (211), (220), and (310) disappeared, and the intensity of peak (110) decreased by bending 135° in NiTi samples. Martensite peaks $\left(1\bar{1}1\right)$, (002), and (111) were observed, indicating the complete transformation from austenite to martensite (Table 2) [6].

Austenite finish (A_f) temperature is defined as the temperature at which phase transformation from martensite to austenite is completed upon heating the alloy. An A_f temperature close to the intraoral temperature should be selected to take advantage of the shape memory effect of NiTi archwires. According to the DSC measurements on commercial NiTi archwires, 0.40 mm × 0.56 mm rectangular Smart SE (Beijing Smart Technology CO., ltd, China), L&H® Titan (Tomy Incorporated, Japan), Damon™ CuNiTi (Ormco, USA) had an A_f close to the oral temperature and possessed shape memory property [11].

The transition temperature of thermally activated NiTi archwires varies widely [12], and the manufacturer should provide the elastic parameters to achieve efficient orthodontic treatment. Moreover, the phase transformation temperatures and mechanical behavior of functionally graded orthodontic archwires were different for the incisive, premolar and molar segments of the same NiTi archwire [13].

2.1. The effects of processing parameters on phase transformations

The effects of thermomechanical treatment conditions on the phase transformation behavior of NiTi archwires were studied previously. Annealing, aging and cold drawing are the treatments for processing NiTi archwires. Direct current heating (DCH) is another processing technique and it can be used for the shape setting of orthodontic archwires. Even a short period of DCH could cause phase transformations [14].

After annealing at 300 °C and 400 °C for 10 min, the density of dislocations was high and maximum resistance to phase transformation from austenite to martensite occurred, resulting in the highest force level of the plateau. By increasing the annealing duration from 10 min to the longer times, the amount of Ni in the matrix decreases during precipitation of Ti_3Ni_4, reducing the load of plateau level. The optimum superelastic properties were obtained for NiTi archwires with an annealing temperature of 300 °C for 30 min. Those parameters provided an A_f temperature lower than body temperature, making the annealing process suitable for orthodontic treatments [15]. When the annealing temperature increased to 500 °C, inhomogenous Ti_3Ni_4 precipitates resulted in a three-step phase transformation that included R-phase in 10 min. As the duration increased to 60 min, inhomogenous precipitation of Ti_3Ni_4 decreased [15].

In a study that used Direct Current (DC), heating was performed at a voltage of 1.6 V and a current of 13.5 A for various short durations [14]. Before application of direct current, as received 50.8 at.% NiTi samples were 15 % cold-rolled. The samples were then heat-treated at 673 K (aged) or 973 K (annealed) for 1.8 × 103 s and then quenched. As-received and annealed samples had a single-step transformation that occurred between austenite and martensite phases. DC heating did not change this behavior. However, a two-step transformation occurred between austenite, R-phase, and martensite in aged samples after direct current heating for 8 s. According to XRD results, besides austenite, R-phase, a small amount of martensite and Ti_3Ni_4 precipitates were detected [14].

Table 2
X-ray diffraction (XRD) planes observed in NiTi alloys[10].

Phase	Major Diffraction Planes	Crystal Structure
Austenitic (B2)	(110), (200), (211)	Body-centered cubic
R-phase	(112), (300), (222), (412), (330)	Rhombohedral
Martensitic (B19′)	(020), $\left(1\bar{1}1\right)$, (002), (111)	Monoclinic

2.2. The effects of stress on phase transformations

Stress is one of the factors that trigger the phase transformations in NiTi archwires. In addition to classical methods for observation of phase transformations, such as DSC and μXRD [8,16], the slope of the electrical resistivity curve in austenite phase and transformation plateau can be used to monitor the transformation from austenite to martensite phase under load [17,18]. Martensite phase was detected after 135° bending in Nitinol™ SE, 35 °C Copper Ni-Ti™ and Neo Sentalloy® samples in the simulated oral environment conditions [8]. The phase transformation of Superelastic Sentalloy® from austenite to martensite at 37 °C started at about 1 mm deflection and continued up to 3.1 mm deflection without any permanent deformation during activation. During deactivation, the complete transformation from martensite to austenite occurred at 0.5 mm of deflection. The superelastic behavior of Sentalloy® tested at 37 °C was seen between 0.5 mm and 2.5 mm in the deactivation. The slope of the deactivation force versus deflection curve was nearly horizontal, indicating superelasticity. Unlike superelastic Sentalloy®, which was not deformed permanently at 3.1 mm of deflection above its A_f temperature, permanent deformation was seen in martensitic Sentalloy® at deflections between 1.0 mm and 1.5 mm [16]. The behavior of shape memory NiTi (Neo Sentalloy® and 35 °C Copper Ni-Ti™) and superelastic NiTi (Nitinol™ Superelastic) in a simulation of oral environment was investigated, and it was reported that all archwires showed martensite phase after bending at 135° and remained stable between 0 and 55 °C [8].

When 5 mm and 10 mm pre-deflections were applied to the NiTi archwires with a composition of 54.65 % of Ni and 45.10 % Ti, as deflection increased, the density of martensite lath decreased, and martensite to austenite transformation was promoted [19].

The applied stress changed the phase transformation temperatures. A_f temperature of Sentalloy® was detected as 22 °C when the sample was unloaded, whereas it shifted to 28 °C when a load that simulates either 1 mm (minimum) and 6 mm (maximum) labiolingual distance was applied. The same experiment was applied to 27 °C Copper Ni-Ti™. An increase in A_f temperature from 31 °C to 33 °C, whereas a decrease in austenite start (A_s) temperature from 16 °C to 4 °C was observed [18].

Generally, DSC is used in experiments to detect the phase transformation temperatures. Another method called bend and free recovery (BFR) was proposed [20] to detect phase transformation temperatures of heat-activated NiTi orthodontic archwires. The method was applied as described in ASTM F 2082; Standard Test Method for Determination of Transformation Temperature of Nickel-Titanium Shape Memory Alloys by Bend and Free Recovery. BFR method provided information about strained orthodontic archwires. Therefore, it was suitable for clinical use to predict the behavior of the archwire in the patient's mouth.

Dynamic mechanical analysis (DMA) is another method for observing A_f temperature. DMA is also considered an efficient method for clinical simulation conditions and can simultaneously provide mechanical and thermal data [21].

2.3. The effects of temperature on phase transformations

DSC was widely used to observe the phase transformations between specific temperature ranges. According to a study focused on the phase transformations of commercial NiTi alloys in a temperature range from −170 °C to 100 °C, the martensitic phase was dominant in nonsuperelastic NiTi samples at room temperature (22 °C) [7]. Small amounts of austenite transformation occurred in the oral environment. On the other hand, the austenite phase was the dominant phase in the oral environment for samples with shape memory characteristics [7]. According to μXRD studies, temperature changes between 0 °C and 55 °C induced phase transformations in shape memory NiTi alloys (Neo Sentalloy® and 35 °C Copper Ni-Ti™). However, minimal change was observed in XRD results for Nitinol™ Superelastic, which had only superelastic property [8].

Formation of new dislocations and interactions with precipitates were claimed for phase transformation changes in the structure of Nitinol™ Heat Activated, 27 °C and 35 °C Copper Ni-Ti™ and Sentalloy® by setting up a thermocycler with the thermocycling effect performing 1000, 5000, and 10,000 cycles within a temperature range between 5 °C and 55 °C [22]. Copper addition in NiTi alloys increased the complexity of the system and formed precipitates, such as $Ti(NiCu_2)$ and Ti_2Cu, which affected phase transformation behavior [22].

3. Surface properties and modification

The major problems encountered in orthodontic treatments with NiTi alloys are corrosion and friction. These two problems result in Ni release and mechanical weakening. Surface modifications are applied to avoid these drawbacks. Surface roughness is one of the main parameters to observe the effects of surface modifications. The measurement methods to investigate surface roughness are classified as destructive and non-destructive techniques. Surface profilometry is a destructive technique. The surface moves as a stylus measures the surface. Optical methods are non-destructive and listed as focus variation microscopy, scanning electron microscopy (SEM) (qualitative analysis), chromatic confocal microscopy, ellipsometry, interferometry, speckle interferometry, angular scattering distributions, laser specular reflectance, and 3D profilogram. Scanning probe microscopy is another non-destructive technique that serves the highest variability. It includes scanning tunneling microscopy (STM), atomic force microscopy (AFM), and magnetic force microscopy (MFM) [23]. The surface roughness (R_a = arithmetical mean of the absolute values) values of eleven different NiTi archwires (Memory Wire, Rematitan® Lite, Titanol® Superelastic, Nitinol™ Superelastic, Nitinol™, Neo Sentalloy®, Ni-Ti®, Orthonol®, Rematitan®, Sentalloy®) varied for measurement method used and variation in batches. The surface roughness values were measured between 0.06 and 1.30 μm for surface profilometry, 0.10 and 0.44 μm for the laser specular reflectance method, and 0.06 and 0.89 μm for AFM [23]. Materials with low surface roughness values are more suitable for applications of orthodontic wires due to low biofilm adhesion, frictional forces between wire and brackets, nickel release rates, and intraoral corrosion caused by wire fracture [24].

The surface topography of NiTi archwires is affected by environment, sterilization technique, recycling, intraoral aging, and type of surface finish. The study investigating the effect of a fluoride-containing medium on surface topography variations of different equiatomic NiTi archwires showed no significant differences in arithmetic mean deviation (ΔR_a) in low fluoride concentrations (in fluoride-containing mouthwash and toothpaste in artificial saliva (<2500 ppm)) in 28 days. On the other hand, ΔR_a values increased from < 70 nm to the range of 120–250 nm in prophylactic gel in saliva containing high amounts of fluoride (17000 ppm) [25]. Moreover, studies in different environments presented no change in surface roughness of Nitinol™ and Titanal™ after 10 h, 6 h, and 10 h of immersion in cold disinfectants; 2 % acidic glutaraldehyde, chlorine dioxide, and iodophor, respectively, according to the specular reflectance power results obtained from laser spectroscopy [26]. The studies comparing different archwires in simulated gastric acid pH 1.2 and 4 showed similar results. Unlike in NiTi samples, significant differences were detected in surface roughness of beta-titanium and stainless steel archwires according to the optical profilometer results [27]. When pH was between 4.8 and 6.6, no significant differences were detected in surface roughness of non-coated, and rhodium-coated and nitrified NiTi [28].

Autoclaving (for 18 min at 134 °C) and cold sterilization (by immersion in 2.45 % glutaraldehyde for 10 h) were used as two different sterilization techniques for esthetic (coated) NiTi archwires [29]. According to AFM analysis results, only the average maximum depth of peak to valley heights (R_z) showed a significant difference between non-sterilized and autoclaved samples. Average overall surface roughness (R_a) and maximum roughness depths (R_q) did not significantly change for none of the samples. In another study [30], no significant differences in surface properties of Neo Sentalloy® were detected after autoclave sterilization.

The recycling effect of NiTi archwires was investigated for three different products (Sentalloy®, Optimalloy, and Ni-Ti®) by 3D profilogram [31]. This method spotted the interferential line formed by the different optical pathways. The size, shape, and surface roughness were analyzed with an image generated by an optical camera. The provided resolution of this method was 0.08 nm, whereas AFM reached a resolution of 0.01 nm in the z direction [32]. According to the 3D profilogram, Ni-Ti® and Optimalloy had higher surface roughness after recycling compared to the control group in the as-received condition, whereas Sentalloy® did not display a significant difference. It was important to note that recycled Ni-Ti® and Optimalloy samples had lower surface roughness than the Sentalloy® control group [31].

The surface roughness of archwires is also affected by intraoral aging. AFM and SEM studies revealed that one month of intraoral aging increased the surface roughness of the non-coated (Sentalloy®, Memory Wire), ion-implanted (Sentalloy® High Esthetic), and polymer-coated (Esthetic Memory Wire, and EverWhite NiTi Cosmetic (American Orthodontics) samples [33]. On the other hand, a clinical use for an average of 91 days and decontamination afterward showed no statistically different surface roughness in heat-activated NiTi archwires, according to the AFM results [34]. Esthetic NiTi archwires were reported to be prone to form a heterogeneous surface containing craters and bumps after clinical use [33]. However, the surface roughness of non-coated NiTi archwires did not change significantly after immersion in many solutions and after clinical use which would be an advantage of NiTi archwires.

Focus variation microscopy is another quantitative technique to analyze surface roughness and generate a topography map by an optical system [35]. However, this technique has low resolution compared to the quantitative methods AFM and a more sensitive method, chromatic confocal microscopy [36].

Surface modification on NiTi archwires was investigated to solve problems related to friction, corrosion, cell and bacteria adhesion, and mechanical properties. Table 3 summarizes the studies based on surface modification of NiTi alloys.

Many coated NiTi archwires are widely used for their esthetic purposes. The most common coating material is epoxy resin due to its electrical insulation, excellent adhesion, chemical resistance, and dimensional stability [37]. The surface roughness of the epoxy-coated sample was found as the highest (R_a = 1.29 ± 0.49), whereas the uncoated TruFlex Nickel Titanium (Ortho Technology) had the

Table 3
Surface modification studies of NiTi alloys.

Technique used	Coating Material Used	Improvement	Ref
Electrodeposition	Co and inorganic fullerene-like tungsten disulfide ($IF-WS_2$) nanoparticle film	66 % of friction coefficient reduction	[48]
Electrodeposition	Bisphenol-A epoxy resin and rutile TiO_2	Increase in surface roughness and contact angle, inhibition of Ni ion release	[39]
Sandblasting	-	Increase in friction forces	[46]
Plasma oxidation with using direct current or radiofrequency	TiO_2	Improvement in superelastic elongation, corrosion resistance and biocompatibility	[42]
Physical Vapor Deposition	TiN/Ti	Protection against mechanical damage, inhibition of Ni ion release	[49]
Aerosol spray method and immersion method	Octadecylphosphonic acid covalently bonded on the oxide covered onto the Nitinol	Stability and corrosion resistance	[50]
Ion beam plating	Diamond-like carbon film	Improvement in wear performance and corrosion resistance	[51]
Plasma electrolytic oxidation	-	Aesthetic appearance and good cytocompatibility	[52]
Chemical deposition	ZnO	21% reduction in the frictional forces and antibacterial activity against *Streptococcus mutans*	[53]
Filtered arching ion plate	TiN	Increase in surface roughness, wettability and biocompatibility	[54]
Electrodeposition	Ag nanoparticles	Decrease in *Streptococcus sanguinis* and *Lactobacillus salivarius* colonies	[55]
High vacuum plasma ion sputtering	Ti	Reduction in surface roughness and friction coefficient Minimum variations in elasticity of modulus and hardness by nanoindentation in artificial saliva	[56]
Ceramic conversion	TiO_2	Improvement in fretting corrosion resistance and reduction in Ni release into the simulated body fluid	[57]

lowest surface roughness (R_a = 0.29 ± 0.16) among the samples of epoxy, polytetrafluoroethylene (PTFE) and rhodium coated NiTi archwires [37]. Similar results were obtained by a non-contact profilometer with a 3-D optical system. In this set-up, epoxy-coated NiTi archwires had the surface roughness R_a as 1.517 ± 0.071 µm, whereas the surface roughness of rhodium-coated samples was 0.297 ± 0.015 µm [38]. Alternatively, the root mean square deviation (R_q) values obtained by AFM are used to measure the surface roughness. The R_q values of conventional NiTi decreased significantly from 578.56 nm to 229.51 nm, 252.22 nm, and 290.64 nm when the surface was modified with titanium, silver, and rhodium, respectively [24]. The epoxy resin (Spectra Epoxy, GAC) and PTFE (Teflon) modified archwires in the esthetic product family, and nitride ion modified (Neo Sentalloy®) surfaces had lower surface roughness values compared to conventional NiTi [24]. Coating with bisphenol-A epoxy resin and rutile TiO_2 with the electrodeposition method increased surface roughness from 0.1 to 0.25 µm and contact angle from 44.4° to 68.6° without affecting shape memory property, bonding strength, microhardness, and superelasticity [39]. A contact angle between 40° and 70° provided the highest cell adhesion in previous studies [40]. The coating inhibited Ni ion release and showed no cytotoxicity on human bone osteosarcoma MG-63 cells. Since surface roughness values vary among different batches and manufacturers, similar products might give inconsistent results.

In addition to coatings, techniques such as plasma oxidation, anodization, and sandblasting have been used for the surface modification of NiTi archwires. Plasma oxidation is a technique used for coating metal surfaces with oxides. It is exerted at a higher potential than anodization, and electrostatic discharge occurs. After the plasma modification, a highly crystalline layer is obtained, which is hard and resistant to corrosion and wear [41]. When radiofrequency was chosen as a power source instead of direct current in the plasma oxidation method, an ideal thin layer (75–100 nm thick) without a nickel-enriched sublayer was obtained. The plasma oxidation increased the hysteresis region, which increased superelastic elongation with a rate of 8.6 % [42]. Anodization is a method similar to plasma oxidation. It uses a direct or alternating current in the presence of an electrolyte solution [43]. Colorful NiTi dental archwires with TiO_2 surface were produced by the anodization method [44]. The thickness of the anodized layer increased with the anodization time. Anodization increased surface roughness, and no nickel release was observed on the surface after anodization [44]. Another method to increase surface roughness is sandblasting. In this method, high pressure steam of abrasive material is used to change the surface texture of a material [45]. The sandblasted samples showed higher friction forces when compared to untreated 0.016-inch nickel-titanium and 0.017 × 0.025-inch Nitinol Classic archwires [46].

Surface-modified NiTi archwires exhibited better corrosion resistance and reduced surface roughness results than conventional NiTi. The highest and lowest surface roughness values (Rq roughness) were obtained for the conventional NiTi and NiTi modified with oxide (Black Diamond,ClassOne Orthodontics), respectively [47]. The other surface-modified NiTi archwires were nitride, Teflon, gold, and epoxy resin-modified samples from the lowest root mean square roughness values.

4. Friction

Friction force plays a crucial role in orthodontic treatments. 50 % or more friction loss can be observed across an archwire [58]. Friction forces that resist sliding depend on the size, shape, material, surface texture of the archwire. The other factors are brackets, ligation of wire in the slot, intraoral factors (such as corrosion, saliva), the direction of force, and distance between the teeth [59].

The studies on measuring friction between archwire and brackets use a special apparatus connected to a universal testing machine [60]. The apparatus models vary and mainly consist of brackets and archwire. The brackets and archwire are aligned to a dental arch form plate. The tensile mode of the universal testing machine is used to pull the one end of the archwire placed in selected brackets [61,62]. In another set-up, two brackets were used. Each was fixed onto half of the metal fixture with archwire ligated on brackets [63]. Set-up with one bracket fixed at different angles between the two metal fixtures, which pulls the archwire to determine the friction force, was applied in.

some studies [64]. The maximum loading under a specific crosshead speed is recorded as the static friction force [61].

Kinetic friction is calculated as the mean frictional force of static peaks at various displacements in a similar set-up [65]. Some studies used a customized apparatus consisting of an extension meter (or micrometer), a strain gauge, a load (or pulse controller), and a bracket-archwire configuration. It is used to detect the resistance levels to sliding [66,67].

Tribometer devices are used to observe the frictional behavior of NiTi archwires. The rotary sliding tribometer measures the friction forces. The stationary part of the equipment bears the archwire and simulated individual teeth. Instead of a bracket system, a stainless steel flat plate is used in the rotary part. The graph of coefficient of friction versus sliding cycles for different loads is obtained [68]. Pin on disc tester is another tribometer device that can measure the friction force by the load cell and the coefficient of friction from the load. The friction is created by a pin sliding on the archwires [69]. Micro-combi tester is another device to measure frictional forces. It uses a Rockwell diamond indenter with a specified tip radius, speed, loading rate, load range, and scratch length. The equipment is interfaced with an integrated optical microscope, an acoustic emission (AE) detection system, and a tangential friction force sensor [56].

Pre-treated archwires were used to avoid friction. Two methods used for pretreatment were mechanical polishing and ion surface treatment [70]. These methods decreased the coefficient of friction significantly when compared to nontreated wires without a significant change in surface roughness [70].

4.1. The effects of brackets on friction

Generally, direct contact between the bracket slot and the archwire results in friction and wear during the orthodontic treatment. The bracket-archwire combination affects the static and kinetic friction results. Compared to conventional brackets, the lower friction forces were reported for stainless steel self-ligating brackets [71,72]. According to the results from the universal testing machine, among stainless steel, beta-titanium, and NiTi archwires, the lowest frictional forces were obtained for NiTi archwires with stainless steel self-ligating brackets with nickel-titanium clips due to the contact with the same material [72].

The static friction coefficient between NiTi archwires (Rematitan® and Activ-Arch, (3 M™ Unitek™, USA)) and stainless steel or titanium brackets were defined as 0.20 for both in the dry state at 34 °C and human saliva at 34 °C. Compared with stainless steel (0.12 for dry state and 0.15 for wet state), the static friction coefficient of NiTi archwires was higher. No difference was detected regarding the bracket type [73].

4.2. The effects of environment on friction

The solutions contacting the archwires and brackets affect the friction forces. According to a study [74] comparing the effect of artificial saliva with pH 6.75 and 0.2 % acidified phosphate fluoride agent on archwires and brackets, frictional resistance increased in fluoride-containing solution. Citric acid, which is used as a food simulating liquid (FSL), increased the static frictional force (gram-force) of Ni-Ti® from 246.98 ± 17.07 to 260.23 ± 16.30 g-forces in 2 weeks and 271.32 ± 16.02 g-forces in 4 weeks of storage. After the 4 weeks of storage of Ni-Ti® in FSL containing ethanol, the static frictional forces increased, whereas there was a decreasing trend in samples stored in FSL containing heptane compared to samples not stored (as-received) [63]. Stainless steel archwires were affected by citric acid and ethanol more when compared to NiTi [63].

The decrease in frictional force was observed when biolubricants such as human saliva, olive oil, Aloe Vera oil, sesame oil, and sunflower oil were applied [75]. The fatty acids in those natural oils provide a lubrication layer between archwires and brackets [75].

5. Corrosion

The oral environment is destructive for dental products due to rapid temperature and pH changes, cavities and microorganisms, high humidity, and the continuous mechanical force that triggers deterioration [76]. Therefore, corrosion resistance is an essential requirement for NiTi archwires. Corrosion resistance is strongly influenced by the composition of the material and the surface roughness [77]. The manufacturing process of NiTi archwires has an important role in its corrosion behavior. SEM analysis and laser scanning confocal microscope system (LSCM) were used to detect the surface morphology after static and electrochemical corrosion of NiTi archwires with different surface characteristics but similar composition. Smooth and dimple surfaces were more resistant to corrosion than a surface with scratches and/or cracks. The residual stress and non-uniform passive film on the surface defects are the main reasons for the high corrosion potential of cracked and scratched surfaces. On the other hand, the number of defects and corrosion were independent of surface roughness [77,78]. Moreover, the present phases of NiTi orthodontic archwires are known to influence the corrosion behavior and Ni ion release. For example, according to a study conducted in artificial saliva for 4 weeks, the austenite phase was more susceptible to corrosion than the martensite phase [79]. Ni ion release was measured for NiTi, stainless steel, ion-implanted NiTi, and copper included NiTi in simulated saliva with pH 5.6–7.0 and temperature 36.5 °C for 7, 14, and 21 days of immersion [80]. The highest Ni ion release values were reported on day 7, and the daily Ni ion release values were measured as 0.93 ± 0.04 μg for NiTi, 0.77 ± 0.05 μg for copper-containing NiTi, 0.67 ± 0.02 μg for ion-implanted NiTi, and 0.66 ± 0.02 μg stainless steel [80]. Although Ni ion released during the whole treatment was only 10 % of the daily dietary intake of Ni ion (200–300 μg), it was important to obtain the lowest Ni ion concentration due to sensitivity to Ni depending on the individual [80]. Ni ion release in saliva with 0.5 $\mu g/cm^2$/week was accepted as the upper limit, and the concentration of leached Ni ions was found to decrease with time [81].

Among the three samples, including stainless steel (ASTM304), beta-titanium alloy, and NiTi archwires, NiTi archwires showed the highest corrosion resistance and consequently the lowest average corrosion current density in artificial saliva [76]. In a previous study [83], an evaluation was made based on changes in surface topography and Ni ion release in artificial saliva (Fusayama-Meyer solution) in the presence of potentiostatic stress. The high tendency of corrosion was detected in Memory Wire and Neo Sentalloy®. On the other hand, Ni-Ti®, Nitinol™, titanium molybdenum, and stainless steel showed low corrosion tendency [83]. The types of corrosion seen in NiTi orthodontic archwires were galvanic, general, and pitting corrosion. Interestingly, the method of brackets manufacturing has been shown to affect the galvanic corrosion rate to an equal or greater extent than the bracket material composition. For example, brazed stainless steel brackets showed the highest galvanic corrosion compared to injection molded stainless steel and titanium brackets [84]. On the other hand, the galvanic current and charge of the couples were not altered significantly by the type of wire: beta-titanium, NiTi, and stainless steel. In terms of pitting susceptibility, 40 °C Copper Ni-Ti™ and 27 °C Copper Ni-Ti™ were more vulnerable to pitting when compared to Ni-Ti® and Nitinol™ Classic. The hysteresis explained this behavior detected only in the cyclic polarization curve of copper-containing NiTi alloys [85]. Nitinol™ was reported to have better resistance to pitting corrosion compared to stainless steel in the Fusayama-Meyer solution [82].

When 0.05 M sodium fluoride salt (NaF) was added to the saliva composition, the type corrosion of SS304 and Nitinol™ wires in this solution was pitting corrosion (Fig. 2) [82]. Pitting and corrosion increased in Ni-Ti® and Optimalloy due to the recycling effect, whereas showed no difference in Sentalloy® [31].

The shape of the NiTi archwires also affected the ion release behavior. According to the study [86], which compared 0.020 in. round and 0.016 × 0.016 in. rectangular NiTi archwires, rectangular NiTi archwires released more Ni and Ti ions in the artificial saliva in 21 days, although the surface areas of both samples were the same [86].

Tribocorrosion is a characteristic that shows the effects of a combination of various mechanical parameters on NiTi orthodontic archwires. It consists of parameters such as corrosion, friction, deformation and mechanical wear. The total wear of a NiTi orthodontic archwire is the combination of those parameters. Since each parameter affects another, it is difficult to predict the exact destructive effects. According to the tribocorrosion tests, the corrosion of NiTi orthodontic archwire increased with the heavy pressure applied on the surface. The potential was lowered by 0.08 V as the mechanical wear load increased from 1 N to 2 N, which showed a stable behavior in terms of the potential and the coefficients of friction.

Moreover, the repassivation of the wire after mechanical wear was fast and detected as 30 s [87]. Auger electron spectroscopy (AES) was a method to detect surface chemistry after tribocorrosion tests. The estimated TiO_2 passive film thickness was 8 nm for NiTi archwires, whereas 1–2 nm of Cr_2O_3 passive film was formed on stainless steel [88]. Total wear and coefficient of friction were higher in Niti archwires than in stainless steel archwires [88]. Ball-on-plate tribocorrosion test results showed that NiTi produced by an additive manufacturing method (laser beam directed energy deposition technique) gave lower open circuit potential, which means better corrosion behavior compared to Ti-6Al-4 V when the mechanical load was applied in PBS [89]. NiTi samples had lower corrosion susceptibility than stainless steel samples for long-term *in vivo* conditions [90].

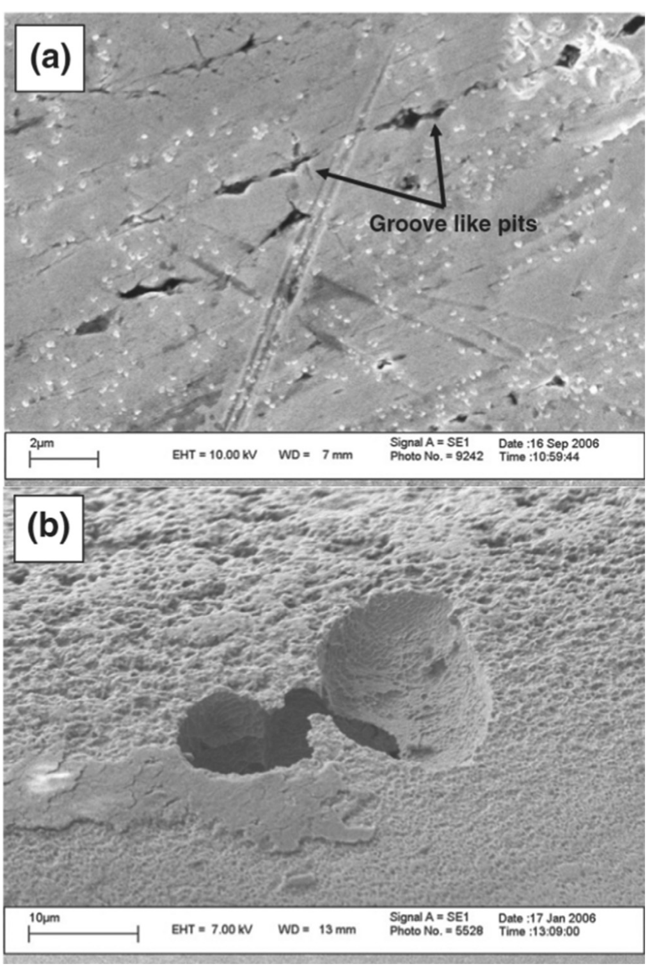

Fig. 2. SEM micrograph of (a) SS304 and (b) Nitinol™ wires after corrosion test in simulated saliva solution containing fluoride ions. Reprinted with permission from [82].

5.1. The effects of environment on corrosion

The oral environment was simulated with a wide range of solutions at different pH values. Simulation with probiotic agents was considered in some studies because of the natural presence of microorganisms in the oral environment.

The decrease in pH of the saliva (in pH 3.9 and 2.5) with lactic acid resulted in an increase in susceptibility to corrosion for NiTi archwires [81]. The studies investigating the surface properties and corrosion resistance of NiTi alloys in an environment of simulated soft drinks at pH 2.5 showed that as the number of defects and surface roughness increased, the corrosion resistance decreased after 28 days of immersion [91]. Corrosion behaviors of Neo Sentalloy®, Nitinol™ Superelastic, Ni-Ti®, and Copper Ni-Ti™ were investigated under different temperatures such as 5, 37, and 60 °C and different pH values such as 3.5, 4.5 and 6 [92]. In the extreme condition, where the pH was 3.5, and the temperature was 60 °C, the release of the elements was the highest.

The effect of fluoride ions added in artificial saliva on the corrosion was observed with breakdown potential. Fluoride ion decreased the breakdown potential in Nitinol™, whereas it showed an increase in breakdown potential in SS304 resulting in pitting corrosion [82].

The surface of Nitinol™ archwires is composed of nickel oxide, which is an unstable oxide, and titanium oxide is dissolved harder than nickel oxide. In phosphate buffered saline (PBS), nickel oxide concentration on the surface initially decreases, and titanium oxide concentration becomes dominant. In a more aggressive environment obtained by the addition of 0.1 % NaF in PBS, Ni ions were released slightly and remained constant for 7 days. Ni release did not decrease, as the aggressive medium did not allow the formation of a stable titanium dioxide layer [93]. The effects of fluoride toothpaste and 1.23 % acidulated phosphate fluoride (APF) on metal ion release were compared. A significant increase in the release of chromium, molybdenum, nickel and iron ions resulted in a decrease in cell viability in (Ni-Ti®) archwires immersed in APF [94]. Compared with beta-titanium, stainless steel and NiTi archwires should be preferred due to more stable behavior in the APF medium [94]. In another study [95], the effects of fluoride prophylactic agents on NiTi and copper added NiTi archwires were compared. According to the SEM micrographs, both acidulated fluoride agent (pH:5.1) and neutral fluoride agent (pH:7) were more corrosive on copper-containing NiTi, when compared to NiTi [95]. When citric acid and ethanol were used as food simulating liquids for Ni-Ti®, citric acid gave a higher corrosion current density than ethanol, while the opposite was the case for corrosion potential. Moreover, NiTi showed the highest passivity and the least negative corrosion current potential in citric acid and ethanol concerning beta-titanium and stainless steel samples [63].

The presence of proteins such as lysozyme, ovalbumin, and bovine serum albumin in saliva inhibited the corrosion of NiTi archwires under mechanical load [96].

5.2. The effects of stress on corrosion

The stress applied to NiTi archwires affected the corrosion behavior. In a study, it was shown that 35 °C Copper Ni-Ti™ alloy in martensite phase at oral temperature released Ni ions due to local stress accumulation and formation of microstructural defects under cyclic loading *in vivo* [97].

The effect of loading force on the dissolution behavior of Nitinol™ and Sentalloy® in artificial saliva regardless of the pH (pH 5.3 and 2) showed that 3 mm deflection increased the Ni ion release significantly [98]. The maximum release was seen between days 0 and 1 and the detected concentration was between 4.3 and 36.8 μg/cm^2. A slight increase in Ni release rate was observed due to protective oxide film deformation on the NiTi surface for stressed samples in 3 days [98]. By mechanical and thermal loading, nickel release increased 10 to 30-fold when NiTi-based archwires were immersed in artificial saliva (pH ~ 5.0) [99].

Tribocorrosion, the degradation process because of the combination of corrosion and wear effect, was observed for Neo Sentalloy®. The mechanical load simulated one day of wear in archwires increased Ni ion release by exceeding the upper limit (0.5 μg/cm^2 per week) by 134 fold. No change in the composition of the surface oxide film on NiTi alloys was detected after the tribological contact in the artificial saliva in 3 weeks [81].

5.3. The effects of temperature on corrosion

The effect of temperature on corrosion was investigated for 40 °C Copper Ni-Ti™, 27 °C Copper Ni-Ti™, Ni-Ti®, and Nitinol™ Classic in artificial saliva at 5, 24, 37 and 45 °C. An increase in corrosion current density was observed as temperature increased for all samples. The threefold increase was detected as the temperature increased from 37 °C to 45 °C in 27 °C Copper Ni-Ti™ and NiTi, which had the lowest A_f temperature [85]. Welding simulation was done in artificial saliva for superelastic NiTi and heat-activated NiTi, and corrosion at the welding joint was investigated [100]. The best results in terms of corrosion potential, current density and polarization resistance were obtained when the welding temperature was at 350 °C [100].

6. Mechanical properties

Desired mechanical properties of an ideal archwire are low stiffness to retain the impact of small forces, high strength to resist permanent deformation, long-term durability without a change in elastic behavior and maximum activation during the orthodontic treatment that lasts over weeks to months. Ease of application in a considerable time scale and low cost should also be considered [1]. Mechanical properties are evaluated by tension, bending, torsion, and deflection tests [1]. Vicker's microhardness test and nanoindentation tests with a Berkovich indenter are used to detect microhardness and nanoindentation hardness, respectively [56,101].

The mechanical properties of NiTi alloys differ according to phase transformations. For instance, high stiffness was obtained by the austenite phase which the elasticity of modulus was between 84 and 98 GPa, and the ultimate tensile strength was 0.84 GPa. On the other hand, the martensite phase provided low stiffness with the elasticity of modulus in the range of 31 to 35 GPa and ultimate tensile strength in the range of 1.4 to 1.7 GPa. Those values were smaller than that of stainless steel (The elasticity of modulus was 200 GPa, and ultimate tensile strength was 2.1 GPa) [102,103]. The yield strength and ultimate tensile strength decreased as the cross-sectional area of the wire increased. According to the three-point and four-point bending test results, the elastic modulus of Nitinol™ was 44.4 GPa for round wires and 33.4 GPa for square and rectangular wires [104]. The Vicker's microhardness of NiTi was detected as 170 ± 1 Vicker's Hardness number (HVN). Copper addition to the structure increased the value to 278 ± 2 HVN. On the other hand, Vicker's hardness of the stainless steel (ASI304) was 484 ± 1 HVN [101].

The constant force is required to provide optimal tooth movement [105]. Therefore, materials with large elastic deformation before breakage and a small modulus of elasticity are preferred [106].

The superelastic property can be defined with three different parameters; the distinctiveness of the superelastic plateau, the deflection at the beginning of the plateau, and the force level of the plateau [107]. The composition and temperature range were the two important properties that affected the presence and absence of a superelastic plateau [108]. The addition of copper to the alloy structure reduced the rigidity, activation, and deactivation moment, and the hysteresis; the distance between the activation and deactivation plateaus [108]. The average temperature of the oral environment is about 36 °C. However, the archwires are subjected to temperatures between 4 °C and 60 °C by hot or cold foods and beverages. Therefore, a wide transformational temperature range is preferable as an archwire to maintain superelastic properties at various temperatures [18].

6.1. The superelastic property

Superelasticity is the elastic response to applied stress. It occurs due to the phase transformation between austenite and martensite crystal structures. NiTi archwires with superelasticity are widely used in orthodontic treatments. The low and constant force is required to provide optimal tooth movement due to the minimal risk of tissue destruction of such forces. On the other hand, hyalinization of the periodontal ligament and root resorption were seen when high-intensity force was applied [109]. Therefore, materials with large elastic deformation before breakage and small modulus of elasticity are preferred [105,106]. The archwires which could produce constant moment at different degrees of rotation in torsion should be preferred. The composition and temperature range were the two important properties that affected the presence and absence of superelastic plateau [108]. 35 °C Copper Ni-Ti™, 27 °C Copper Ni-Ti™, Elastinol™ 35 (Masel®, USA), NiTi (Morelli® Orthodontia, Brasil), Nitinol™ Heat-activated and Neo Sentalloy® had plateau corresponding to constant moment. Adding copper to the alloy structure reduced the rigidity, activation and deactivation moment. Moreover, the hysteresis, the distance between the activation and deactivation plateaus, was reduced [108].

The superelastic ratio was used as an indicator of superelasticity. It is defined as the slope of the specified data point with respect to the slope of the start point of the unloading curve in the force/displacement graph. Sterilization decreased the superelastic ratio and applied force of 0.016-inch diameter superelastic NiTi archwires (Force I (American Orthodontics, USA), Rematitan® Lite, and Trueform™ (G&H® Orthodontics, USA)) during loading and unloading phases. The effect of dry heat sterilization was more significant than steam sterilization in both measured properties. The reason for the difference was explained by the higher temperature and longer duration in dry sterilization (160 °C, 120 min) when compared to steam sterilization (121 °C, 24 min, 15 psi) [110].

6.2. The effects of deflection on mechanical properties

The load–deflection characteristics are important to determine the levels of force applied to the tooth. For the cases of the mild to moderately crowded tooth arrangement, the activation and deactivation forces should be low. According to the study which tested commercial products in terms of activation and deactivation under deflection of 0.2 mm to 2 mm, five superelastic samples with dimensions of 0.017 × 0.025 inch^2, NiTi (Morelli® Orthodontia, Brasil), Nitinol™ Heat Activated, NeoSentalloy® F200, Elastinol™ 35 (Masel®, USA), 35 °C Copper Ni-Ti™ showed deactivation moment between 190 and 350 g which was suitable for mild to moderately crowded tooth treatments [111]. Among thermo-active Copper Ni-Ti™ archwires (27 °C, 35 °C and 40 °C), 40 °C Copper Ni-Ti™ archwire had the lowest deactivation forces at 5 mm, 4 mm, and 3 mm of activation [112]. A three-point bending test with 2 and 4 mm deflection showed that thermal NiTi exerted lower mean load values than superelastic NiTi archwires did during the unloading phase [113]. Moreover, the mean force during the unloading phase increased as the cross-section of the archwire increased. The superelastic archwires had a steep and narrow hysteresis curve, whereas thermal archwires had a wide interval between loading and unloading forces for the samples with the same size and deflection [113]. A broader plateau was obtained in 4 mm deflection compared to 2 mm deflection due to the stress-induced transformation property of archwires (Fig. 3) [113].

6.3. The effects of heat treatment and aging on mechanical properties

The heat treatment is applied for bending that has effects on mechanical properties of NiTi alloy since the crystalline structure of NiTi alloy changes with bending. In chair-side clinical practice, bending is classified into two types as cold bending and heat forming. Heat forming applications are made by electrical current and laboratory furnace. Three-point bending test results on commercial products Titanol® Low Force, Neo Sentalloy® F80, and 35 °C Copper Ni-Ti™ showed that the cold bending technique gave predictable and slightly lower bending forces at 1.5 mm deflection when compared to untreated samples. On the other hand, the samples heated by electrical current showed unpredictable results, whereas the samples heated by a laboratory furnace had stable results when treated for only 2 s at 550 °C [115]. The effects of direct current heating on tensile deformation behavior were observed for annealed, aged, and as received NiTi samples. According to the results, the superelastic hysteresis loop of as-received samples changed slightly up to 25 s of heating [14]. After 25 s, stress plateaus decreased with time. On the other hand, annealed samples showed a gradual increase in stress plateau up to 25 s [14]. A dramatic drop was observed after 100 s of heating, and the stress plateau started to increase.

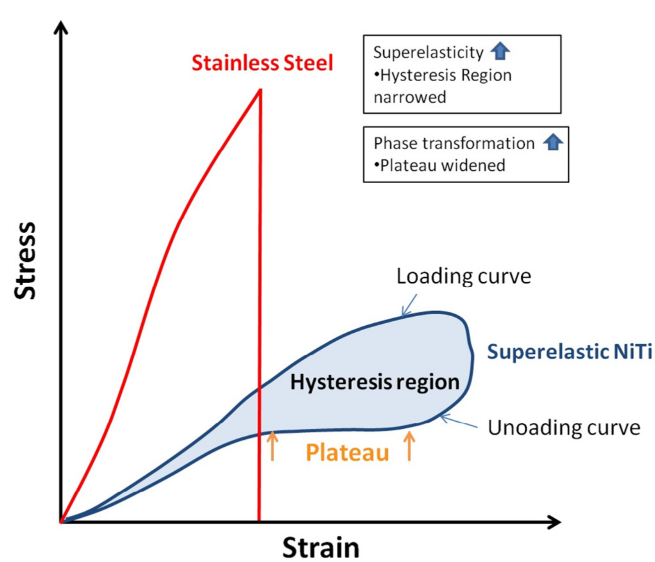

Fig. 3. Stress vs Strain curve of NiTi archwire and comparison with Stainless Steel archwires, . Adapted from [114]

The aging of orthodontic archwires changed the superelastic properties of the NiTi orthodontic archwires. The effects of intraoral aging and sterilization on NiTi and copper-containing NiTi archwires were studied. According to the results, 8 weeks of intraoral aging and autoclave sterilization decreased the deactivation mean load in most deflections (from 2.00 mm to 0.00 with 0.2 mm intervals) for NiTi samples and all deflections for copper included NiTi samples [116]. The strain recovery in Nitinol™ Heat-activated archwires enhanced after aging for up to 30 min at 400 °C and the superelasticity reached a value close to as-received samples [117]. 60 min of aging decreased the strain recovery [117].

The temperature range was 5.6–58.5 °C at the incisor and 7.9–54 °C at the premolar based on the study on Asian and Caucasian samples. The temperatures between 33 and 37 °C were observed in 79 % and 92 % of the 24 h duration at the incisor and premolar site, respectively [118]. The short-term applications of hot (80 °C) and cold (10 °C) water were observed to simulate the oral environment. The temperature changes strongly affected the bending force at constant strain exerted by rectangular Neo Sentalloy® and 35 °C Copper Ni-Ti™, whereas work-hardened conventional NiTi was minimally affected [119]. No superelastic behavior was observed at room temperature for aged samples. With direct current heating, aged samples become perfectly superelastic at 10th second [14]. The heat treatment of 500 °C for 5 h in air resulted in an increase in tensile strength of Elgiloy® (Co-Cr-Ni superalloy) (Elgiloy Specialty Metals, IL, USA) samples from the range between 1.4 and 2.1 GPa to the range between 1.6 and 2.8 GPa [120]. Generally, heat treatment is used to increase the tensile strength of the archwire. Since the variation of the results was high, the prediction of tensile strength after heat treatment of as-received Elgiloy® archwires became a challenge for orthodontists [120].

6.4. The effects of environment on mechanical properties

The environmental effect on mechanical properties and lifetime of NiTi archwires were studied and compared with stainless steel archwires [121]. The samples were tested in air, distilled water, citric acid in water with pH 3.5, NaCl in water, Fusayama-Meyer artificial saliva, and fluoridated artificial saliva. DMA results showed that stainless steel archwires had a longer lifetime than Sentalloy®, 27 °C Copper Ni-Ti™, and 35 °C Copper Ni-Ti™ except for the Sentalloy® in air, and 27 °C Copper Ni-Ti™ (rectangular) in NaCl solution [121]. The lifetime of Sentalloy® decreased sharply in all immersion solutions whereas, the decrease in the stainless steel lifetime was distinctive for the presence of inorganic salts, artificial and fluoridated saliva [121]. In another study [95], the effects of fluoride prophylactic agents, such as acidulated type Phos-flur gel (pH: 5.1) and neutral type Prevident 5000 (pH: 7), on the elasticity of modulus and yield strength of NiTi and copper-containing NiTi were investigated in loading and unloading conditions. Unlike the copper-containing NiTi, NiTi samples showed a significant decrease in the elasticity of modulus and yield strength in unloading condition [95]. The effects of disinfectants such as 2 % acidic glutaraldehyde, chlorine dioxide, and iodophor were tested for 10 h, 6 h, and 10 h respectively on Nitinol™ and Titanal™ by bending and tensile tests. No differences were detected in stiffness, strength, or surface topography of the wires [26].

7. Biological properties

Based on the clinical information claimed that high nickel ion concentration induced hypersensitivity in some patients, *in vitro* tests for biological properties of NiTi archwires were studied to

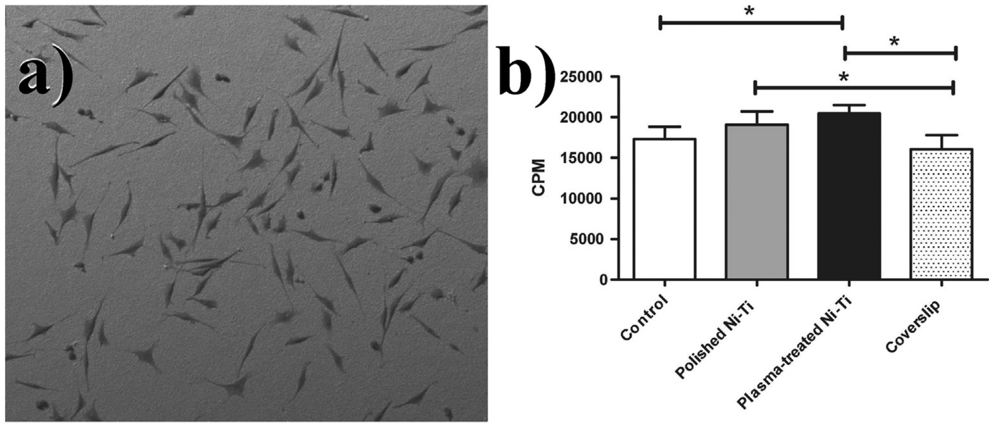

Fig. 4. **(a)** SEM image of L929 cells cultured for 24 h on Nitinol™ and **(b)** the proliferation of human PBMNCs *in vitro*. *$p < 0.05$ compared to the indicated bar. . Reproduced with permission from [127]

observe the effects [122]. The biological properties of NiTi alloys were tested in terms of cytotoxicity, sensitization, and genotoxicity.

The biocompatibility of NiTi archwires was observed by a method based on static immersion in artificial saliva and EDX measurements. According to the ex-situ experiment results, the acidic environment became dominant in a short period after immersion. At later stages of immersion (30 days), a protein-based protective film was formed on the surface of the samples [123]. The protein-based protective film led to the formation of carbon-rich pile-ups observed on clinical samples [123].

In terms of oxidative stress and cell viability, NiTi yielded the lowest cell viability index (94.13 ± 1.23) in the sample set of stainless steel, nickel-titanium (Titanol®), copper-nickel-titanium (Copper Ni-Ti™), rhodium-coated nickel-titanium (BioForce® High Esthetic), cobalt-chromium blue Elgiloy® (Coballoy, Dentsply GAC, USA), and titanium-molybdenum [124]. The experiments on mouse fibroblast L929 cells using 8-hydroxy-2-deoxyguanosine in DNA as an oxidative stress marker showed that nickel-titanium (Titanol®) produced the highest, whereas titanium-molybdenum and stainless steel generated the lowest oxidative stress. Furthermore, nickel-titanium (Titanol®) significantly gave the lowest cell viability and the highest inhibition of cell growth in the sample set with the exceptions of titanium-molybdenum for cell viability and cobalt-chromium Blue Elgiloy® for inhibition of cell growth [124]. Equiatomic NiTi alloy had no cytotoxic effect on human skin fibroblasts (PK84) in three days. No growth inhibition, cell lysis, intracellular granulation, and morphological changes were observed after the end-point dilution minimal essential medium (MEM) extract test was applied [125]. Cell proliferation, viability, and cytoskeletal organization of human epithelial embryonic cells (L132) and human embryonic palatal mesenchymal cells (HEPM) were observed on Nitinol™ [126]. Cell proliferation rates were calculated as 73 % for HEPM and 79 % for L132 cell lines. The cytoskeleton of HEPM cells was observed with actin labeling, and highly spread cells with overloaded cytoskeletons were observed [126].

Peripheral blood mononuclear cells (PBMNCs) and L929 cells were used to evaluate the cytocompatibility of Nitinol™, which was oxidized by using different procedures to improve biocompatibility and prevent the release of Ni [127]. After 24 h of incubation, L929 cells adhered and spread well onto the surface of Nitinol™ samples (see Fig. 4a). Nitinol™ samples modified with plasma have also been reported to stimulate proliferation of human PBMNCs, possibly due to their adhesion-promoting properties. The *in vitro* neuro-cytotoxicity of NiTi, copper-containing NiTi, and stainless steel was tested by lactate dehydrogenase release assay [128]. The assay was done on murine cortical cells after 24 h of interaction with the NiTi archwires. Among those three types, stainless steel was significantly neuro-cytotoxic. The cell death on stainless steel samples stemmed from free radicals on the surface [128].

The effects of nickel concentration in artificial saliva on human peripheral blood mononuclear cells derived from nickel sensitive and non-sensitive individuals were compared for Bioforce®, stainless steel, and nitrogen implanted NiTi (Sentalloy® IonGuard) archwires. The detected nickel ion concentration was between 0.4 and 4.1 ppb. The highest release was observed for Bioforce® and in the case of cyclic straining. Despite this, the maximum released concentration of nickel was 700 times lower than the cytotoxicity limit of human peripheral blood mononuclear cells [122]. The success of the NiTi alloy on biological responses is due to its good corrosion resistance and limitation in ion release. However, the cytotoxicity, genotoxicity, and sensitization tests encompassed an extended period should be considered due to long periods of orthodontic treatments [125].

Genotoxicity is another critical property for product safety. Salmonella reverse mutation test revealed that the equiatomic NiTi alloy did not show any genotoxicity in 2 days [125]. Another study used the Micronucleus technique to observe genotoxicity [129]. In this technique, chromosomal loss and breakage can be detected by identifying blocked binucleated cells during cytokinesis due to a microfilament assembly inhibitor, cytochalasin-B (Cyt-B) [130].

Probiotic bacteria *Lactobacillus reuteri Prodentis* in saliva used for simulation of an oral environment of a patient with poor oral hygiene was increased the surface roughness of uncoated NiTi compared to the conditions with saliva only [131]. There is a positive correlation between surface roughness and bacterial adhesion [37]. According to the study based on a comparison of epoxy, tetrafluoroethylene and rhodium coated NiTi archwires, epoxy-coated samples had the highest surface roughness. The adhesion of *Streptococcus mutans* and *Streptococcus sobrinus* increased compared to the uncoated samples [37]. When copper was added to NiTi alloy, the adhesion of *Streptococcus mutans*, surface energy, and surface roughness values increased [132].

8. Clinical perspectives

Orthodontic treatments are classified into three stages; leveling/alignment, correction of molar relationship and space closure, and the last stage; finishing and retention [1]. NiTi archwires are early-stage solutions for correcting rotation, tipping, leveling, and torquing [2]. Reasonable control on sufficient magnitude of the force is provided due to the low load to deflection ratio of NiTi alloys. This property makes NiTi alloys a good candidate for archwires in the first stage (leveling/alignment) of the treatment [5]. However, in later phases of the orthodontic treatment, NiTi alloys are not preferred due to low deformability [5]. In the main stage, the correction of molar relationship and space closure, open coil springs and closing loops made of NiTi alloys are used in addition to NiTi archwires. For the finishing and retention stage, NiTi archwires are not recommended due to their poor formability which results in difficulty in the control of individual teeth [1]. Moreover, the detected interlot variations in transition temperature range and force delivery resulted in failures in orthodontic treatments [133].

The studies aimed at investigating pain perception showed no statistical difference between conventional NiTi, and superelastic NiTi archwire (Sentalloy) after the first four hours [134].

8.1. Mechanical properties in clinical studies

Irreversible tissue damage and root resorption were triggered by the high magnitude of force. Therefore, a low and constant force is required for optimal tooth movement [103]. It provides a constant stress in the periodontal ligament and minimal tissue damage [109]. The superelastic property of NiTi alloys provided the constant force and included stress-induced phase transformations between austenite to martensite by forming a plateau region. After the deactivation, a reverse phase transformation from martensite to austenite formed a second plateau with a lower force when compared to the first plateau (Fig. 3). In the second plateau region, the shape change was observed to obtain a constant force. A practitioner activated the archwire and obtained the plateau region, and the constant force was achieved [103].

Compared with single-stranded Sentalloy® in the round form, the coaxial tubular superelastic NiTi archwires were more efficient in the treatment of relieving anterior mandibular crowding [135]. The 0.016 in. coaxial tubular superelastic NiTi archwires were also superior to 0.016 in. single-stranded superelastic Rematitan® Lite wire in the treatment of the lower anterior region in 4, 8, and 12 weeks [136]. When compared to 0.0155-inch diameter multiple-stranded stainless steel wire, 0.014-inch diameter superelastic nickel-titanium alloy wire (Ni-Ti®) gave improved results in the treatment of lower labial segment in 6 weeks of treatment, although the alignment of crowding did not differ [137].

A_f temperature, as an indicator of phase transformation, has clinical importance for thermally activated NiTi archwires. The archwires with A_f close to the oral temperature (35–37 °C) are selected to benefit from the shape memory effect [11]. When heat-activated Copper Ni-Ti™ archwires are deformed at room temperature, they retain the deformed shape at that temperature and return their original shape at body temperature. The highly crowded teeth are aligned by cooling the heat-activated Copper Ni-Ti™ archwires. At body temperature, heat-activated Copper Ni-Ti™ archwires return their original shape and align teeth in their new position [138]. The fractures occurred on NiTi archwires with a percentage of 1.7 and in the posterior region with a percentage of 76 [139].

8.2. Biofilm formation

The possible consequences of long-term usage of NiTi archwires are biofilm formation and calcification, which increase surface roughness and porosity. From a clinical perspective, it resulted in friction variance, altered torque expression, and suppressed superelastic properties [140]. The study based on retrieved NiTi archwires from different manufacturers revealed that the formation of proteinaceous biofilm mainly consisted of amide, alcohol, and carbonate was observed as well as NaCl, KCl and Ca-P precipitates in 6 months [141]. As the duration of intraoral exposure increased, a reduction in grain size, delamination, pitting, and crevice corrosion was seen [141]. *Streptococcus mutans* and *Lactobacillus* colonization on uncoated (Memory Wire) and epoxy resin coated (EverWhite NiTi Cosmetic) NiTi archwires were compared [142]. According to the results, epoxy-coated NiTi archwires showed higher Lactobacillus colonization when compared to uncoated NiTi archwires due to the peeling of the coating and higher surface roughness. *Streptococcus mutans* showed an increase in saliva for both archwire applications [142].

8.3. Ni ion release

Another clinically significant parameter was toxicity due to the release of Ni ions. Petoumenou et al. [143] measured Ni ion concentration in the saliva of the patients. The results obtained immediately after the implementation showed that bands and brackets increased Ni concentration in saliva from 34 μg/L (before treatment) to 78 μg/L, whereas the implementation of NiTi archwires increased Ni concentration to 56 μg/L [143]. The leaching and concentration of Ni in saliva decreased in 10 weeks [143]. The clinical study comparing the salivary nickel concentrations showed that copper-containing NiTi released more Ni ions when compared to epoxy coated samples, whereas Ni ion concentration was lower compared to conventional NiTi [144]. Clinical studies on 115 retrieved archwires (Titanol® 'Low Force' and 'Martensitic', 35 °C Copper Ni-Ti™, nickel-titanium from Ortho Organizers® Inc, USA) from different companies showed that none of the archwires had corrosion in one month [145]. As the treatment period increased to 8 weeks, corrosion was seen in 19 % of the archwires. The percentage increased to 50 % in 24 weeks, and for the treatment period longer than 36 weeks, it reached 91 % [145].

According to the tribocorrosion studies, the repassivation rate of NiTi orthodontic archwires is fast [87]. Long-term *in vivo* conditions showed that NiTi archwires had better corrosion properties than stainless steel [90]. Moreover, under mechanical wear, NiTi archwires had stable potential and coefficient of friction [87]. Those properties decreased the risk of Ni ion release and increased the biocompatibility of NiTi archwires in clinical studies.

9. Computational studies

Computational studies are used to predict the mechanical behavior of NiTi orthodontic archwires in specific conditions. Finite element modeling claimed by the experimental studies is a widely used method to predict the mechanical performance of NiTi archwires. The effects of intraoral temperature differences on the forces generated by NiTi orthodontic archwires were investigated by 3D finite element models by using a Gibbs-potential-based formulation and thermodynamic principles [146]. The superelastic tensile and bending test model gave consistent results with the experimental data. Moreover, the dental displacement can be predicted by finite element numerical simulations for NiTi superelastic shape memory alloys. Numerical simulations revealed the sensitivity of the mechanical response to the temperature, the importance of bone stiffness in the force generated by the wire to correct the misalignment, and the importance of constant and continuous force on bone remodeling [147]. The effect of hydrogen diffusion on the thermomechanical performance of NiTi shape memory archwires was also modeled by finite element software due to the risk of embrittlement, wire fractures, and loss in superelasticity by hydrogen in titanium alloy systems [148,149]. In another study, the bending force response at different wire-bracket friction coefficients and sliding friction force were modeled for superelastic

NiTi archwires by the 3D finite element method [150,151]. According to the results, at contact friction coefficients higher than 0.4, the deactivation force was ineffective and suppressed the tooth movement [151].

Force and stress level calculations in a model of crowded lower front teeth in reduced periodontium were done by finite element method for nonsuperelastic NiTi, superelastic NiTi and stainless steel archwires. The effect of material variation in a specific orthodontic case was discussed. The results were compared to a physiological periodontal model [152].

Welding is a promising technique to obtain orthodontic archwires with different characteristics. Laser welding can be used to obtain customized orthodontic archwires which exert specified forces on misaligned teeth. This procedure lowers the treatment time. The finite element method was used to predict optimum laser welding parameters such as laser power and velocity for an effective weld [153]. The best method was selected by using artificial neural network and training algorithms. A machine learning model with a multi layered feed forward neural network used the Ni ion release data in the literature to predict the best composition of NiTi shape memory alloys for dental applications [154]. According to the results, 51.5 % Ni in the shape memory alloy composition is optimum, with the lowest amount of Ni ion released into the oral cavity [154]. In another study, CALPHAD quaternary phase analysis was applied to detect intermetallic compounds and phases in the weld zone between NiTi and austenitic stainless steel [155].

10. Future perspectives

According to the analysis presented in the introduction part of this review, computational studies gained importance in the last five years. The planning stage should be organized well to decrease the treatment time and avoid unpredicted tooth movements. In this scope, computational programs that specified the route of the treatment became important. In terms of material improvement, the studies subjected to Ni release and decrease in bacterial activities may lead to the future of the area. For example, Liu et. al. obtained an antibacterial surface and inhibited Ni ion release by surface modification after etching with HNO_3 at different time intervals. They enabled Ni dissolution and the formation of TiO_2 nanolayer on the surface. Treating with 1H,1H,2H,2Hperfluorodecyltrimethoxysilane (PFDMS) increased the contact angle [156]. A clinical study revealed that TiO_2 coating on the surface decreased the surface roughness and Streptococcus mutans adhesion. However, at the end of one month, the advantage of coated archwire on surface roughness was lost [157]. Electrodeposition of silver ions on NiTi archwires decreased the *Streptococcus sanguinis* and *Lactobacillus salivarius* colonies by 90 % after 1 day of incubation [55]. Obtaining a coating that endures to the end of a specified period of the orthodontic treatment without deformation and has antibacterial properties will support NiTi archwires to be a good candidate for orthodontic archwires. In terms of mechanical properties, welding of different alloys to form composite archwire offers a solution for various treatments without changing the archwires in short intervals [158].

11. Conclusion

A wide range of NiTi archwires, which are now widely used in the clinic, are available from many manufacturers, including coated esthetic wires. Structural properties of NiTi archwires were reviewed mainly by considering the phase transformations, and the effects of processing parameters such as aging and annealing, and stress and temperature were explained. The surface modification studies were applied generally to reduce the Ni ion release and corrosion. The corrosion and friction behavior of NiTi archwires were widely studied since those were the primary source of deformation. Biocompatibility has been evaluated in the reviewed studies, mainly by cytotoxicity, sensitivity, and genotoxicity tests. Bacterial adhesion for *Streptococcus mutans*, *Streptococcus sobrinus*, and *Lactobacillus* was observed in studies by comparison to other archwire types such as stainless steel and coated NiTi archwires. According to clinical studies, Ni release does not rise to toxic levels, and although it initially increases, it decreases significantly after 10 weeks of use. The NiTi archwires were generally used at the leveling and alignment stage, and batch and brand-based differences increased the margin of error in the orthodontic treatment. Welding and coating technologies that can improve the material and the computational methods that specify treatment routes will provide reliable, short, and cost-effective treatments in combination.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

References

[1] I. Sifakakis, C. Bourauel, in: Nickel–titanium products in daily orthodontic practice, Elsevier Ltd, 2017, https://doi.org/10.1016/b978-0-08-100383-1.00006-0.
[2] N. Pandis, C.P. Bourauel, Nickel-Titanium (NiTi) arch wires: The clinical significance of super elasticity, Semin. Orthod. 16 (2010) 249–257, https://doi.org/10.1053/j.sodo.2010.06.003.
[3] D. Rose, A. Quick, M. Swain, P. Herbison, Moment-to-force characteristics of preactivated nickel-titanium and titanium-molybdenum alloy symmetrical T-loops, Am. J. Orthod. Dentofac. Orthop. 135 (2009) 757–763, https://doi.org/10.1016/j.ajodo.2007.06.015.
[4] O. Barwart, The effect of temperature change on the load value of Japanese NiTi coil springs in the superelastic range, Am. J. Orthod. Dentofacial Orthop. 110 (1996) 553–558, https://doi.org/10.1016/S0889-5406(96)70065-3.
[5] H.P. Chang, Y.C. Tseng, A novel β-titanium alloy orthodontic wire, Kaohsiung, J Med. Sci. 34 (2018) 202–206, https://doi.org/10.1016/j.kjms.2018.01.010.
[6] M. Iijima, H. Ohno, I. Kawashima, K. Endo, W.A. Brantley, I. Mizoguchi, Micro X-ray diffraction study of superelastic nickel-titanium orthodontic wires at different temperatures and stresses, Biomaterials 23 (2002) 1769–1774, https://doi.org/10.1016/S0142-9612(01)00303-9.
[7] T.G. Bradley, W.A. Brantley, B.M. Culbertson, Differential scanning calorimetry (DSC) analyses of superelastic and nonsuperelastic nickel-titanium orthodontic wires, Am. J. Orthod. Dentofacial Orthop. 109 (1996) 589–597, https://doi.org/10.1016/S0889-5406(96)70070-7.
[8] M. Iijima, W.A. Brantley, I. Kawashima, H. Ohno, W. Guo, Y. Yonekura, I. Mizoguchi, Micro-X-ray diffraction observation of nickel-titanium orthodontic wires in simulated oral environment, Biomaterials 25 (2004) 171–176, https://doi.org/10.1016/S0142-9612(03)00473-3.
[9] W.A. Brantley, M. Iijima, T.H. Grentzer, Temperature-modulated DSC study of phase transformations in nickel-titanium orthodontic wires, Thermochim Acta 392–393 (2002) 329–337, https://doi.org/10.1016/S0040-6031(02)00119-3.
[10] M. Iijima, W.A. Brantley, W.H. Guo, W.A.T. Clark, T. Yuasa, I. Mizoguchi, X-ray diffraction study of low-temperature phase transformations in nickel-titanium orthodontic wires, Dent. Mater. 24 (2008) 1454–1460, https://doi.org/10.1016/j.dental.2008.03.005.
[11] C.C. Ren, Y.X. Bai, H.M. Wang, Y.F. Zheng, S. Li, Phase transformation analysis of varied nickel-titanium orthodontic wires, Chin. Med. J. (Engl) 121 (2008) 2060–2064, https://doi.org/10.1097/00029330-200810020-00022.
[12] T.S. Spini, F.P. Valarelli, R.H. Cançado, K.M.S.d. Freitas, D.J. Villarinho, Transition temperature range of thermally activated nickel-titanium archwires, J. Appl. Oral Sci. 22 (2) (2014) 109–117.
[13] P.F. Rodrigues, F.M.B. Fernandes, R. Magalhães, E. Camacho, A. Lopes, A.S. Paula, R. Basu, N. Schell, Thermo-mechanical characterization of NiTi orthodontic archwires with graded actuating forces, J. Mech. Behav. Biomed. Mater. 107 (2020) 103747.
[14] Y.B. Wang, Y.F. Zheng, Y. Liu, Effect of short-time direct current heating on phase transformation and superelasticity of Ti-50.8at.%Ni alloy, J. Alloy. Compd. 477 (2009) 764–767, https://doi.org/10.1016/j.jallcom.2008.10.131.
[15] S.M. Seyyed Aghamiri, M. Nili Ahmadabadi, H. Shahmir, F. Naghdi, S. Raygan, Study of thermomechanical treatment on mechanical-induced phase

transformation of NiTi and TiNiCu wires, J. Mech. Behav. Biomed. Mater. 21 (2013) 32–36, https://doi.org/10.1016/j.jmbbm.2013.01.014.
[16] N. Segal, J. Hell, D.W. Berzins, Influence of stress and phase on corrosion of a superelastic nickel-titanium orthodontic wire, Am. J. Orthod. Dentofac. Orthop. 135 (2009) 764–770, https://doi.org/10.1016/j.ajodo.2007.04.042.
[17] J. Ferčec, I. Anžel, R. Rudolf, Stress dependent electrical resistivity of orthodontic wire from the shape memory alloy NiTi, Mater. Des. 55 (2014) 699–706, https://doi.org/10.1016/j.matdes.2013.10.041.
[18] M. Santoro, D.N. Beshers, Nickel-titanium alloys: Stress-related temperature transitional range, Am. J. Orthod. Dentofac. Orthop. 118 (2000) 685–692, https://doi.org/10.1067/mod.2000.98113.
[19] J. Jafari, S.M. Zebarjad, S.A. Sajjadi, Effect of pre-strain on microstructure of Ni-Ti orthodontic archwires, Mater. Sci. Eng., A 473 (2008) 42–48, https://doi.org/10.1016/j.msea.2007.03.067.
[20] N.A. Obaisi, M.T.S. Galang-Boquiren, C.A. Evans, T.G.P. Tsay, G. Viana, D. Berzins, S. Megremis, Comparison of the transformation temperatures of heat-activated Nickel-Titanium orthodontic archwires by two different techniques, Dent. Mater. 32 (2016) 879–888, https://doi.org/10.1016/j.dental.2016.03.017.
[21] R.P. Kusy, J.Q. Whitley, Thermal and mechanical characteristics of stainless steel, titanium-molybdenum, and nickel-titanium archwires, Am. J. Orthod. Dentofac. Orthop. 131 (2007) 229–237, https://doi.org/10.1016/j.ajodo.2005.05.054.
[22] D.W. Berzins, H.W. Roberts, Phase transformation changes in thermocycled nickel-titanium orthodontic wires, Dent. Mater. 26 (2010) 666–674, https://doi.org/10.1016/j.dental.2010.03.010.
[23] C. Bourauel, T. Fries, D. Drescher, R. Plietsch, Surface roughness of orthodontic wires via atomic force microscopy, laser specular reflectance, and profilometry, Eur. J. Orthod. 20 (1998) 79–92, https://doi.org/10.1093/ejo/20.1.79.
[24] M. Krishnan, S. Seema, B. Tiwari, H.S. Sharma, S. Londhe, V. Arora, Surface Characterization of Nickel Titanium Orthodontic Arch Wires, Med. J. Armed Forces India. 71 (2015) S340-S345.e5. https://doi.org/10.1016/j.mjafi.2013.12.006.
[25] H.H. Huang, Variation in surface topography of different NiTi orthodontic archwires in various commercial fluoride-containing environments, Dent. Mater. 23 (2007) 24–33, https://doi.org/10.1016/j.dental.2005.11.042.
[26] J.E. Buckthal, R.P. Kusy, Effects of cold disinfectants on the mechanical properllies and the surface topography of nickel-titanium arch wires, Am. J. Orthod. Dentofac. Orthop. 94 (1988) 117–122, https://doi.org/10.1016/0889-5406(88)90359-9.
[27] L. Baidas, H. Alkawari, N. Alshihah, G. Almashaan, H. Alwaalan, Impact of Simulated Gastric Acid on Surface Roughness and Frictional Resistance of Orthodontic Archwires: An In vitro Study, J. Clin. Diagnostic Res. 14 (2020) 10–15, https://doi.org/10.7860/jcdr/2020/45051.14297.
[28] Z. Jusufi Osmani, B. Poljšak, S. Zelenika, E. Kamenar, K. Marković, M. Perčić, V. Katić, Ion Release and Surface Changes of Nickel-Titanium Archwires Induced by Changes in the pH Value of the Saliva—Significance for Human Health Risk Assessment, Materials (Basel). 15 (6) (2022) 1994.
[29] M. Shamohammadi, E. Hormozi, M. Moradinezhad, M. Moradi, M. Skini, V. Rakhshan, Surface topography of plain nickel-titanium (NiTi), as-received aesthetic (coated) NiTi, and aesthetic NiTi archwires sterilized by autoclaving or glutaraldehyde immersion: A profilometry/SEM/AFM study, Int. Orthod. 17 (2019) 60–72, https://doi.org/10.1016/j.ortho.2019.01.016.
[30] C. Pernier, B. Grosgogeat, L. Ponsonnet, G. Benay, M. Lissac, Influence of autoclave sterilization on the surface parameters and mechanical properties of six orthodontic wires, Eur. J. Orthod. 27 (2005) 72–81, https://doi.org/10.1093/ejo/cjh076.
[31] Y. Il Chang, S.H. Lee, Effects of recycling on the mechanical properties and the surface topography of nickel-titanium alloy wires, Am. J. Orthod. Dentofac. Orthop. 120 (2001) 654–663, https://doi.org/10.1067/mod.2001.118997.
[32] V. Sudarsan, Materials for Hostile Chemical Environments, Elsevier Inc., 2017. https://doi.org/10.1016/B978-0-12-801300-7.00004-8.
[33] R. Rongo, G. Ametrano, A. Gloria, G. Spagnuolo, A. Galeotti, S. Paduano, R. Valletta, V. D'Antò, Effects of intraoral aging on surface properties of coated nickel-titanium archwires, Angle Orthod. 84 (4) (2014) 665–672.
[34] J.P. Alcock, M.E. Barbour, J.R. Sandy, A.J. Ireland, Nanoindentation of orthodontic archwires: The effect of decontamination and clinical use on hardness, elastic modulus and surface roughness, Dent. Mater. 25 (2009) 1039–1043, https://doi.org/10.1016/j.dental.2009.03.003.
[35] K. Schmeidl, M. Wieczorowski, K. Grocholewicz, M. Mendak, J. Janiszewska-Olszowska, Frictional properties of the tinbtazro orthodontic wire—a laboratory comparison to popular archwires, Materials (Basel). 14 (21) (2021) 6233.
[36] C. Jumelle, A. Hamri, G. Egaud, C. Mauclair, S. Reynaud, V. Dumas, S. Pereira, T. Garcin, P. Gain, G. Thuret, Comparison of four methods of surface roughness assessment of corneal stromal bed after lamellar cutting, Biomed. Opt. Express 8 (2017) 4974, https://doi.org/10.1364/boe.8.004974.
[37] M.A. Asiry, I. AlShahrani, S. Almoammar, B.H. Durgesh, A.A. Al Kheraif, M.I. Hashem, Influence of epoxy, polytetrafluoroethylene (PTFE) and rhodium surface coatings on surface roughness, nano-mechanical properties and biofilm adhesion of nickel titanium (Ni-Ti) archwires, Mater. Res. Express 5 (2) (2018) 026511.
[38] J. Alsanea, H. Al Shehri, Evaluation of nanomechanical properties, surface roughness, and color stability of esthetic nickel-titanium orthodontic archwires, J. Int. Soc. Prev. Community Dent. 9 (2019) 33, https://doi.org/10.4103/jispcd.JISPCD_365_18.
[39] J.L. Xu, T. Lai, J.M. Luo, Preparation and characterization of the aesthetic coating on nickel-titanium orthodontic archwire by electrophoretic deposition, Prog. Org. Coatings. 137 (2019) 105271.
[40] X.M. Liu, S.L. Wu, P.K. Chu, C.Y. Chung, C.L. Chu, Y.L. Chan, K.O. Lam, K.W.K. Yeung, W.W. Lu, K.M.C. Cheung, K.D.K. Luk, Nano-scale surface morphology, wettability and osteoblast adhesion on nitrogen plasma-implanted NiTi shape memory alloy, J. Nanosci. Nanotechnol. 9 (2009) 3449–3454, https://doi.org/10.1166/jnn.2009.NS15.
[41] R.P.H. Chang, Plasma Oxidation of Semiconductor and Metal Surfaces., Elsevier B.V., 1983. https://doi.org/10.1016/b978-0-444-42252-1.50070-9.
[42] D.A. Campeol, C.P. Fontoura, M.M. Rodrigues, C. Aguzzoli, Assessment of mechanical and corrosion properties of plasma oxidized medical grade NiTi wire, Vacuum 171 (2020), https://doi.org/10.1016/j.vacuum.2019.109013 109013.
[43] S. Ebnesajjad, Surface Preparation of Metals, Elsevier Inc., 2010. https://doi.org/10.1016/B978-1-4377-4461-3.10006-9.
[44] C.-L. Yang, F.-L. Chen, S.-W. Chen, Anodization of the dental arch wires, Mater. Chem. Phys. 100 (2-3) (2006) 268–274.
[45] S.N. Papageorgiou, N. Pandis, Clinical evidence on orthodontic bond failure and associated factors, Elsevier Ltd, 2017. https://doi.org/10.1016/b978-0-08-100383-1.00012-6.
[46] M. Karasawa, T. Tsumura, K. Fujita, M. Ito, S. Nagasawa, K. Yamada, Study on the frictional properties between bracket and wire by sandblast processing, Orthod. Waves. 74 (2015) 48–53, https://doi.org/10.1016/j.odw.2015.02.001.
[47] M. Krishnan, S. Seema, A.V. Kumar, N.P. Varthini, K. Sukumaran, V.R. Pawar, V. Arora, Corrosion resistance of surface modified nickel titanium archwires, Angle Orthod. 84 (2) (2014) 358–367.
[48] G.R. Samorodnitzky-Naveh, M. Redlich, L. Rapoport, Y. Feldman, R. Tenne, Research Article Inorganic fullerene-like tungsten disulfide nanocoating for friction reduction of nickel – titanium alloys R esearch A rticle, Nanomedicine. 4 (8) (2009) 943–950.
[49] J.K. Liu, I.H. Liu, C. Liu, C.J. Chang, K.C. Kung, Y.T. Liu, T.M. Lee, J.L. Jou, Effect of titanium nitride/titanium coatings on the stress corrosion of nickel-titanium orthodontic archwires in artificial saliva, Appl. Surf. Sci. 317 (2014) 974–981, https://doi.org/10.1016/j.apsusc.2014.08.132.
[50] I. Milošev, M. Metikoš-Huković, Ž. Petrović, Influence of preparation methods on the properties of self-assembled films of octadecylphosphonate on Nitinol: XPS and EIS studies, Mater. Sci. Eng., C 32 (2012) 2604–2616, https://doi.org/10.1016/j.msec.2012.08.010.
[51] S. Kobayashi, Y. Ohgoe, K. Ozeki, K. Sato, T. Sumiya, K.K. Hirakuri, H. Aoki, Diamond-like carbon coatings on orthodontic archwires, Diam. Relat. Mater. 14 (2005) 1094–1097, https://doi.org/10.1016/j.diamond.2004.11.036.
[52] O. Jung, J.-P. Becker, R. Smeets, M. Gosau, G. Becker, B. Kahl-Nieke, A.-K. Jung, M. Heiland, A. Kopp, M. Barbeck, T. Koehne, Surface characteristics of esthetic nickel-titanium and beta-titanium orthodontic archwires produced by plasma electrolytic oxidation (PEO)-primary results, Materials (Basel). 12 (9) (2019) 1403.
[53] M. Kachoei, A. Nourian, B. Divband, Z. Kachoei, S. Shirazi, Zinc-oxide nanocoating for improvement of the antibacterial and frictional behavior of nickel-titanium alloy, Nanomedicine. 11 (19) (2016) 2511–2527.
[54] S. Jin, Y. Zhang, Q. Wang, D. Zhang, S. Zhang, Influence of TiN coating on the biocompatibility of medical NiTi alloy, Colloids Surfaces B Biointerfaces. 101 (2013) 343–349, https://doi.org/10.1016/j.colsurfb.2012.06.029.
[55] F.J. Gil, E. Espinar-Escalona, N. Clusellas, J. Fernandez-Bozal, M. Artes-Ribas, A. Puigdollers, New bactericide orthodonthic archwire: NiTi with silver nanoparticles, Metals (Basel). 10 (2020) 1–12, https://doi.org/10.3390/met10060702.
[56] P. Anuradha, N.K.S. Varma, A. Balakrishnan, Reliability performance of titanium sputter coated Ni-Ti arch wires: Mechanical performance and nickel release evaluation, Biomed. Mater. Eng. 26 (2015) 67–77, https://doi.org/10.3233/BME-151550.
[57] H. Dong, X. Ju, H. Yang, L. Qian, Z. Zhou, Effect of ceramic conversion treatments on the surface damage and nickel ion release of NiTi alloys under fretting corrosion conditions, J. Mater. Sci. - Mater. Med. 19 (2008) 937–946, https://doi.org/10.1007/s10856-007-3222-3.
[58] C. Bourauel, P. Husmann, N. Höse, L. Keilig, A. Jäger, Die Friktion bei der bogengeführten Zahnbewegung - Eine Übersicht, Informationen Aus Orthod. & Kieferorthopädie. 39 (2007) 18–26. https://doi.org/10.1055/s-2007-960546.
[59] M. Redlich, R. Tenne, Nanoparticle coating of orthodontic appliances for friction reduction, Elsevier Inc., 2019. https://doi.org/10.1016/B978-0-12-815886-9.00013-9.
[60] V. Arash, K. Anoush, S.M. Rabieer, M. Rahmatei, S. Tavanafar, The effects of silver coating on friction coefficient and shear bond strength of steel orthodontic brackets, Scanning. 37 (2015) 294–299, https://doi.org/10.1002/sca.21212.
[61] M. Murayama, Y. Namura, T. Tamura, H. Iwai, N. Shimizu, Relationship between friction force and orthodontic force at the leveling stage using a coated wire, J. Appl. Oral Sci. 21 (2013) 554–559, https://doi.org/10.1590/1679-775720130325.
[62] B. Patil, N.S. Patil, V.V. Kerudi, S.S. Chitko, A.R. Maheshwari, H.A. Patil, N.P. Pekhale, P.D. Tekale, Friction between archwire of different sizes, cross section, alloy and brackets ligated with different brands of low friction elastic

ligatures-an invitro study, J. Clin. Diagnostic Res. 10 (2016) ZC18–ZC22. https://doi.org/10.7860/JCDR/2016/16769.7534.
[63] D.I. Sherief, N.H. Abbas, The effect of food simulating liquids on the static frictional forces and corrosion activity of different types of orthodontic wires, J. World Fed. Orthod. 6 (2017) 165–170, https://doi.org/10.1016/j.ejwf.2017.11.002.
[64] Y. Kim, J.Y. Cha, C.J. Hwang, H.S. Yu, S.G. Tahk, Comparison of frictional forces between aesthetic orthodontic coated wires and self-ligation brackets, Korean, J. Orthod. 44 (2014) 157–167, https://doi.org/10.4041/kjod.2014.44.4.157.
[65] T. Stefański, A. Kloc-Ptaszna, L. Postek-Stefańska, The effect of simulated erosive conditions on the frictional behavior of different orthodontic bracket-wire combinations, Dent. Med. Probl. 56 (2019) 173–177, https://doi.org/10.17219/dmp/105832.
[66] Y.C. Liaw, Y.Y.M. Su, Y.L. Lai, S.Y. Lee, Stiffness and frictional resistance of a superelastic nickel-titanium orthodontic wire with low-stress hysteresis, Am. J. Orthod. Dentofac. Orthop. 131 (578) (2007) e12–578.e18, https://doi.org/10.1016/j.ajodo.2006.08.015.
[67] Y. Yanase, H. Ioi, M. Nishioka, I. Takahashi, Effects of sliding velocity on friction an in vitro study at extremely low sliding velocity approximating orthodontic tooth movement, Angle Orthod. 84 (2014) 451–458, https://doi.org/10.2319/060513-427.1.
[68] I. Ben Naceur, khaled Elleuch, Tribological properties of deflected NiTi superelastic archwire using a new experimental set-up: Stress-induced martensitic transformation effect, Tribol. Int. 146 (2020), https://doi.org/10.1016/j.triboint.2019.106033 106033.
[69] M.N. Ahmad, A.S. Mahmud, M.F. Razali, N. Mokhtar, Force-deflection behaviour of NiTi archwires in a polytetrafluoroethylene (Teflon) bracket system, Materwiss, Werksttech 50 (2019) 289–294, https://doi.org/10.1002/mawe.201800225.
[70] A. Wichelhaus, M. Geserick, R. Hibst, F.G. Sander, The effect of surface treatment and clinical use on friction in NiTi orthodontic wires, Dent. Mater. 21 (2005) 938–945, https://doi.org/10.1016/j.dental.2004.11.011.
[71] V. Cacciafesta, M.F. Sfondrini, A. Ricciardi, A. Scribante, C. Klersy, F. Auricchio, Evaluation of friction of stainless steel and esthetic self-ligating brackets in various bracket-archwire combinations, Am. J. Orthod. Dentofac. Orthop. 124 (2003) 395–402, https://doi.org/10.1016/S0889-5406(03)00504-3.
[72] M.R.G. Monteiro, L.E.d. Silva, C.N. Elias, O.d.V. Vilella, Frictional resistance of self-ligating versus conventional brackets in different bracketarchwire-angle combinations, J. Appl. Oral Sci. 22 (3) (2014) 228–234.
[73] R.P. Kusy, J.Q. Whitley, W.W. Ambrose, J.G. Newman, Evaluation of titanium brackets for orthodontic treatment: Part I. The passive configuration, Am. J. Orthod. Dentofac. Orthop. 114 (1998) 558–572, https://doi.org/10.1016/s0889-5406(98)70176-3.
[74] C.T. Kao, S.J. Ding, C.K. Wang, H. He, M.Y. Chou, T.H. Huang, Comparison of frictional resistance after immersion of metal brackets and orthodontic wires in a fluoride-containing prophylactic agent, Am. J. Orthod. Dentofac. Orthop. 130 (568) (2006) e1–568.e9, https://doi.org/10.1016/j.ajodo.2005.09.028.
[75] A. Dridi, W. Bensalah, S. Mezlini, S. Tobji, M. Zidi, Influence of bio-lubricants on the orthodontic friction, J. Mech. Behav. Biomed. Mater. 60 (2016) 1–7.
[76] K. Małkiewicz, M. Sztogryn, M. Mikulewicz, A. Wielgus, J. Kamiński, T. Wierzchoń, Comparative assessment of the corrosion process of orthodontic archwires made of stainless steel, titanium–molybdenum and nickel–titanium alloys, Arch. Civ. Mech. Eng. 18 (2018) 941–947, https://doi.org/10.1016/j.acme.2018.01.017.
[77] V.J. Pulikkottil, S. Chidambaram, P.U. Bejoy, P.K. Femin, P. Paul, M. Rishad, Corrosion resistance of stainless steel, nickel-titanium, titanium molybdenum alloy, and ion-implanted titanium molybdenum alloy archwires in acidic fluoride-containing artificial saliva: An in vitro study, J. Pharm. Bioallied Sci. (2016), https://doi.org/10.4103/0975-7406.192032.
[78] C. Abalos, A. Paúl, A. Mendoza, E. Solano, F.J. Gil, Influence of topographical features on the fluoride corrosion of Ni-Ti orthodontic archwires, J. Mater. Sci. - Mater. Med. 22 (2011) 2813–2821, https://doi.org/10.1007/s10856-011-4460-y.
[79] J. Briceño, A. Romeu, E. Espinar, J.M. Llamas, F.J. Gil, Influence of the microstructure on electrochemical corrosion and nickel release in NiTi orthodontic archwires, Mater. Sci. Eng., C 33 (2013) 4989–4993, https://doi.org/10.1016/j.msec.2013.08.024.
[80] A. Charles, P. Gangurde, S. Jacob, S. Jatol-Tekade, R.S. Senkutvan, V. Vadgaonkar, Evaluation of nickel ion release from various orthodontic arch wires: An in vitro study, J. Int. Soc. Prev. Community Dent. 4 (1) (2014) 12.
[81] P. Močnik, T. Kosec, J. Kovač, M. Bizjak, The effect of pH, fluoride and tribocorrosion on the surface properties of dental archwires, Mater. Sci. Eng., C 78 (2017) 682–689, https://doi.org/10.1016/j.msec.2017.04.050.
[82] M. Mirjalili, M. Momeni, N. Ebrahimi, M.H. Moayed, Comparative study on corrosion behaviour of Nitinol and stainless steel orthodontic wires in simulated saliva solution in presence of fluoride ions, Mater. Sci. Eng., C 33 (2013) 2084–2093, https://doi.org/10.1016/j.msec.2013.01.026.
[83] F. Widu, D. Drescher, R. Junker, C. Bourauel, Corrosion and biocompatibility of orthodontic wires, J. Mater. Sci. - Mater. Med. 10 (1999) 275–281, https://doi.org/10.1023/A:1008953412622.
[84] A. Bakhtari, T.G. Bradley, W.K. Lobb, D.W. Berzins, Galvanic corrosion between various combinations of orthodontic brackets and archwires, Am. J. Orthod. Dentofac. Orthop. 140 (2011) 25–31, https://doi.org/10.1016/j.ajodo.2010.05.021.
[85] D.K. Pun, D.W. Berzins, Corrosion behavior of shape memory, superelastic, and nonsuperelastic nickel-titanium-based orthodontic wires at various temperatures, Dent. Mater. 24 (2008) 221–227, https://doi.org/10.1016/j.dental.2007.05.003.
[86] A. Azizi, A. Jamilian, F. Nucci, Z. Kamali, N. Hosseinikhoo, L. Perillo, Release of metal ions from round and rectangular NiTi wires, Prog. Orthod. 17 (2016), https://doi.org/10.1186/s40510-016-0123-3.
[87] P. Mocnik, T. Kosec, Tribo-corrosion properties of a NiTi dental wire, Mater. Tehnol. 48 (2014) 467–472.
[88] T. Kosec, P. Močnik, U. Mezeg, A. Legat, M. Ovsenik, M. Jenko, J.T. Grant, J. Primožič, Tribocorrosive Study of New and In Vivo Exposed Nickel Titanium and Stainless Steel Orthodontic Archwires, Coatings. 10 (2020) 230, https://doi.org/10.3390/coatings10030230.
[89] M. Buciumeanu, A. Bagheri, F.S. Silva, B. Henriques, A.F. Lasagni, N. Shamsaei, Tribocorrosion Behavior of NiTi Biomedical Alloy Processed by an Additive Manufacturing Laser Beam Directed Energy Deposition Technique, Materials (Basel). 15 (2) (2022) 691.
[90] J. Primozic, M. Hren, U. Mezeg, A. Legat, Tribocorrosion Susceptibility and Mechanical Characteristics of As-Received and Long-Term In-Vivo Aged Nickel-Titanium and Stainless-Steel Archwires, Materials (Basel). 15 (4) (2022) 1427.
[91] C. Abalos, A. Paul, A. Mendoza, E. Solano, C. Palazon, F.J. Gil, Influence of soft drinks with low pH on different Ni-Ti orthodontic archwire surface patterns, J. Mater. Eng. Perform. 22 (2013) 759–766, https://doi.org/10.1007/s11665-012-0311-3.
[92] H.S. Ahn, M.J. Kim, H.J. Seol, J.H. Lee, H. Il Kim, Y.H. Kwon, Effect of pH and temperature on orthodontic NiTi wires immersed in acidic fluoride solution, J. Biomed. Mater. Res. - Part B Appl. Biomater. 79 (2006) 7–15, https://doi.org/10.1002/jbm.b.30505.
[93] M. Cioffi, D. Gilliland, G. Ceccone, R. Chiesa, A. Cigada, Electrochemical release testing of nickel-titanium orthodontic wires in artificial saliva using thin layer activation, Acta Biomater. 1 (2005) 717–724, https://doi.org/10.1016/j.actbio.2005.07.008.
[94] T. Yanisarapan, P. Thunyakitpisal, P. on Chantarawaratit, Corrosion of metal orthodontic brackets and archwires caused by fluoride-containing products: Cytotoxicity, metal ion release and surface roughness, Orthod. Waves. 77 (2018) 79–89, https://doi.org/10.1016/j.odw.2018.02.001.
[95] M.P. Walker, R.J. White, K.S. Kula, Effect of fluoride prophylactic agents on the mechanical properties of nickel-titanium-based orthodontic wires, Am. J. Orthod. Dentofac. Orthop. 127 (2005) 662–669, https://doi.org/10.1016/j.ajodo.2005.01.015.
[96] C. Zhang, L. He, Y. Chen, D. Dai, Y. Su, L. Shao, Corrosion Behavior and in Vitro Cytotoxicity of Ni-Ti and Stainless Steel Arch Wires Exposed to Lysozyme, Ovalbumin, and Bovine Serum Albumin, ACS, Omega. 5 (2020) 18995–19003, https://doi.org/10.1021/acsomega.0c02312.
[97] W.A. Brantley, W. Guo, W.A.T. Clark, M. Iijima, Microstructural studies of 35 °C Copper Ni-Ti orthodontic wire and TEM confirmation of low-temperature martensite transformation, Dent. Mater. 24 (2008) 204–210, https://doi.org/10.1016/j.dental.2007.04.004.
[98] J.K. Liu, T.M. Lee, I.H. Liu, Effect of loading force on the dissolution behavior and surface properties of nickel-titanium orthodontic archwires in artificial saliva, Am. J. Orthod. Dentofac. Orthop. 140 (2011) 166–176, https://doi.org/10.1016/j.ajodo.2010.03.031.
[99] M. Arndt, A. Brück, T. Scully, A. Jäger, C. Bourauel, Nickel ion release from orthodontic NiTi wires under simulation of realistic in-situ conditions, J. Mater. Sci. 40 (2005) 3659–3667, https://doi.org/10.1007/s10853-005-0448-7.
[100] N.C.M. Lia Fook, J.D. Costa, M.B. Sousa, J.J.N. Alves, C.J. Araújo, A.R.N. Campos, R.A.C. Santana, Evaluation of the corrosion resistance of welded Ni-Ti wires for orthodontic use, Mater. Today:. Proc. 14 (2019) 678–685, https://doi.org/10.1016/j.matpr.2019.02.006.
[101] M.V. Alfonso, E. Espinar, J.M. Llamas, E. Rupérez, J.M. Manero, J.M. Barrera, E. Solano, F.J. Gil, Friction coefficients and wear rates of different orthodontic archwires in artificial saliva, J. Mater. Sci. - Mater. Med. 24 (2013) 1327–1332, https://doi.org/10.1007/s10856-013-4887-4.
[102] R.P. Kusy, A.R. Greenberg, Effects of composition and cross section on the elastic properties of orthodontic wires, Angle Orthod. 51 (1981) 325–341, https://doi.org/10.1043/0003-3219(1981)051<0325:EOCACS>2.0.CO;2.
[103] R.P. Kusy, A review of contemporary archwires: Their properties and characteristics, Angle Orthod. 67 (1997) 197–208, https://doi.org/10.1043/0003-3219(1997)067<0197:AROCAT>2.3.CO;2.
[104] R.P. Kusy, A.M. Stush, Geometric and material parameters of a nickel-titanium and a beta titanium orthodontic arch wire alloy, Dent. Mater. 3 (1987) 207–217, https://doi.org/10.1016/S0109-5641(87)80035-0.
[105] P.D. Wilkinson, P.S. Dysart, J.A.A. Hood, G.P. Herbison, Load-deflection characteristics of superelastic nickel-titanium orthodontic wires, Am. J. Orthod. Dentofac. Orthop. 121 (2002) 483–495, https://doi.org/10.1067/mod.2002.121819.
[106] S.E. Muraviev, G.B. Ospanova, M.Y. Shlyakhova, Estimation of force produced by nickel-titanium superelastic archwires at large deflections, Am. J. Orthod. Dentofac. Orthop. 119 (2001) 604–609, https://doi.org/10.1067/mod.2001.114538.
[107] D. Segner, D. Ibe, Properties of superelastic wires and their relevance to orthodontic treatment, Eur. J. Orthod. 17 (1995) 395–402, https://doi.org/10.1093/ejo/17.5.395.

[108] J. De, A. Gurgel, S. Kerr, J.M. Powers, A. Pinzan, Torsional properties of commercial nickel-titanium wires during activation and deactivation, Am. J. Orthod. Dentofac. Orthop. 120 (2001) 76–79, https://doi.org/10.1067/mod.2001.115147.

[109] R.C.L. Sachdeva, S. Miyazaki, Superelastic Ni-Ti Alloys in Orthodontics, Butterworth-Heinemann Ltd (1990), https://doi.org/10.1016/b978-0-7506-1009-4.50042-1.

[110] S. Alavi, S.H. Raji, A.A. Ghorbani, Effects of steam and dry-heat sterilization on bending properties of NiTi wires, Orthod. Waves. 68 (2009) 123–128, https://doi.org/10.1016/j.odw.2009.01.005.

[111] J.d.A. Gurgel, S. Kerr, J.M. Powers, V. LeCrone, Force-deflection properties of superelastic nickel-titanium archwires, Am. J. Orthod. Dentofac. Orthop. 120 (4) (2001) 378–382.

[112] D.C. Mallory, J.D. English, J.M. Powers, W.A. Brantley, H.I. Bussa, Force-deflection comparison of superelastic nickel-titanium archwires, Am. J. Orthod. Dentofac. Orthop. 126 (2004) 110–112, https://doi.org/10.1016/j.ajodo.2004.03.012.

[113] E. Gatto, G. Matarese, G. Di Bella, R. Nucera, C. Borsellino, G. Cordasco, Load-defl ection characteristics of superelastic and thermal nickel-titanium wires, Eur. J. Orthod. 35 (2013) 115–123, https://doi.org/10.1093/ejo/cjr103.

[114] M.d.A. Ferreira, M.A. Luersen, P.C. Borges, Nickel-titanium alloys: a systematic review, Dental Press, J. Orthod. 17 (3) (2012) 71–82.

[115] L.M. Brauchli, H. Keller, C. Senn, A. Wichelhaus, Influence of bending mode on the mechanical properties of nickel-titanium archwires and correlation to differential scanning calorimetry measurements, Am. J. Orthod. Dentofac. Orthop. 139 (2011) e449–e454, https://doi.org/10.1016/j.ajodo.2009.12.034.

[116] H. Zarif Najafi, S.R. Gavareshki, Evaluating the effect of clinical usage and autoclave sterilization on the load deflection properties of three different orthodontic wires: Ex-vivo study, Int. Orthod. 17 (2019) 469–477, https://doi.org/10.1016/j.ortho.2019.06.007.

[117] S.M. Seyyed Aghamiri, M. Nili Ahmadabadi, S. Raygan, I. Haririan, M.S. Ahmad Akhondi, The mechanical and thermal behaviors of heat-treated Ni-Rich NiTi orthodontic archwires, J. Mater. Eng. Perform. 18 (2009) 843–847, https://doi.org/10.1007/s11665-009-9489-4.

[118] R.J. Moore, J.T.F. Watts, J.A.A. Hood, D.J. Burritt, Intra-oral temperature variation over 24 hours, Eur. J. Orthod. 21 (1999) 249–261, https://doi.org/10.1093/ejo/21.3.249.

[119] T.R. Meling, J. Ødegaard, The effect of short-term temperature changes on superelastic nickel-titanium archwires activated in orthodontic bending, Am. J. Orthod. Dentofac. Orthop. 119 (2001) 263–273, https://doi.org/10.1067/mod.2001.112451.

[120] S.M. Philip, B.W. Darvell, Effect of heat treatment on the tensile strength of 'Elgiloy' orthodontic wire, Dent. Mater. 32 (2016) 1036–1041, https://doi.org/10.1016/j.dental.2016.05.011.

[121] O. Prymak, A. Klocke, B. Kahl-Nieke, M. Epple, Fatigue of orthodontic nickel-titanium (NiTi) wires in different fluids under constant mechanical stress, Mater. Sci. Eng., A 378 (2004) 110–114, https://doi.org/10.1016/j.msea.2003.10.332.

[122] W. Jia, M.W. Beatty, R.A. Reinhardt, T.M. Petro, D.M. Cohen, C.R. Maze, E.A. Strom, M. Hoffman, Nickel release from orthodontic arch wires and cellular immune response to various nickel concentrations, J. Biomed. Mater. Res. 48 (1999) 488–495, https://doi.org/10.1002/(SICI)1097-4636(1999)48:4<488::AID-JBM14>3.0.CO;2-D.

[123] S.M. Toker, D. Canadinc, Evaluation of the biocompatibility of NiTi dental wires: A comparison of laboratory experiments and clinical conditions, Mater. Sci. Eng., C 40 (2014) 142–147, https://doi.org/10.1016/j.msec.2014.03.060.

[124] S. Spalj, M. Mlacovic Zrinski, V. Tudor Spalj, Z. Ivankovic Buljan, In-vitro assessment of oxidative stress generated by orthodontic archwires, Am. J. Orthod. Dentofac. Orthop. 141 (2012) 583–589, https://doi.org/10.1016/j.ajodo.2011.11.020.

[125] D.J. Wever, A.G. Veldhuizen, M.M. Sanders, J.M. Schakenraad, J.R. Van Horn, Cytotoxic, allergic and genotoxic activity of a nickel-titanium alloy, Biomaterials 18 (1997) 1115–1120, https://doi.org/10.1016/S0142-9612(97)00041-0.

[126] L. El Medawar, P. Rocher, J.C. Hornez, M. Traisnel, J. Breme, H.F. Hildebrand, Electrochemical and cytocompatibility assessment of NiTiNOL memory shape alloy for orthodontic use, Biomol. Eng. 19 (2002) 153–160, https://doi.org/10.1016/S1389-0344(02)00041-2.

[127] M. Jenko, M. Godec, A. Kocijan, R. Rudolf, D. Dolinar, M. Ovsenik, M. Gorenšek, R. Zaplotnik, M. Mozetic, A new route to biocompatible Nitinol based on a rapid treatment with H 2 /O 2 gaseous plasma, Appl. Surf. Sci. 473 (2019) 976–984, https://doi.org/10.1016/j.apsusc.2018.12.140.

[128] A. David, D. Lobner, In vitro cytotoxicity of orthodontic archwires in cortical cell cultures, Eur. J. Orthod. 26 (2004) 421–426, https://doi.org/10.1093/ejo/26.4.421.

[129] O. Erdem, S. Çetinkaya, M. Kaplan, E. Çırak, S.M. Gökçe, C. Akay, Investigation of the genotoxic effect and trace element levels of the nickel-titanium arc-wires used in orthodontic treatment, Toxicol. Lett. 258 (2016) S297, https://doi.org/10.1016/j.toxlet.2016.06.2030.

[130] M. Fenech, The in vitro micronucleus technique, Mutat. Res. - Fundam. Mol. Mech. Mutagen. 455 (2000) 81–95, https://doi.org/10.1016/S0027-5107(00)00065-8.

[131] I. Musa Trolic, Z. Todoric, D. Pop Acev, P. Makreski, B. Pejova, S. Spalj, Effects of the presence of probiotic bacteria in the aging medium on the surface roughness and chemical composition of two dental alloys, Microsc. Res. Tech. 82 (2019) 1384–1391, https://doi.org/10.1002/jemt.23290.

[132] K.S. Abraham, N. Jagdish, V. Kailasam, S. Padmanabhan, Streptococcus mutans adhesion on nickel titanium (NiTi) and copper-NiTi archwires: A comparative prospective clinical study, Angle Orthod. 87 (2017) 448–454, https://doi.org/10.2319/040516-270.1.

[133] R.C. Pompei-Reynolds, G. Kanavakis, Interlot variations of transition temperature range and force delivery in copper-nickel-titanium orthodontic wires, Am. J. Orthod. Dentofac. Orthop. 146 (2014) 215–226, https://doi.org/10.1016/j.ajodo.2014.05.017.

[134] L.M. Fernandes, B. Ogaard, L. Skoglund, Pain and discomfort experienced after placement of a conventional or a superelastic NiTi aligning archwire. A randomized clinical trial., J. Orofac. Orthop. 59 (1998) 331–339. https://doi.org/10.1007/BF01299769.

[135] J. Joseph, VivekSuku Ninan, MerinElsa Abraham, J. John, KarunKoshy Cherian, ReemaMary Thomas, Arch expansion efficiency of coaxial tubular superelastic nickel-titanium in comparison to single-stranded superelastic nickel-titanium while relieving mandibular anterior crowding: A randomized controlled study, J. Int. Soc. Prev. Community Dent. 9 (1) (2019) 60.

[136] B. Sebastiana, Alignment efficiency of superelastic coaxial nickel-titanium vs superelastic single-stranded nickel-titanium in relieving mandibular anterior crowding a randomized controlled prospective study, Angle Orthod. 82 (2012) 703–708, https://doi.org/10.2319/072111-460.1.

[137] A.E. West, M.L. Jones, R.G. Newcombe, Multiflex versus superelastic: A randomized clinical trial of the tooth alignment ability of initial arch wires, Am. J. Orthod. Dentofac. Orthop. 108 (1995) 464–471, https://doi.org/10.1016/S0889-5406(95)70046-3.

[138] B. Aydın, N.E. Şenışık, Ö. Koşkan, Evaluation of the alignment efficiency of nickeltitanium and copper-nickel-titanium archwires in patients undergoing orthodontic treatment over a 12-week period: A single-center, randomized controlled clinical trial, Korean, J. Orthod. 48 (2018) 153–162, https://doi.org/10.4041/kjod.2018.48.3.153.

[139] U. Guzman, L. Jerrold, A. Abdelkarim, An in vivo study on the incidence and location of fracture in round orthodontic archwires, J. Orthod. 40 (2013) 307–312, https://doi.org/10.1179/1465313313Y.0000000062.

[140] T. Eliades, C. Bourauel, Intraoral aging of orthodontic materials: The picture we miss and its clinical relevance, Am. J. Orthod. Dentofac. Orthop. 127 (2005) 403–412, https://doi.org/10.1016/j.ajodo.2004.09.015.

[141] T. Eliades, G. Eliades, A.E. Athanasiou, T.G. Bradley, Surface characterization of retrieved NiTi orthodontic archwires, Eur. J. Orthod. 22 (2000) 317–326, https://doi.org/10.1093/ejo/22.3.317.

[142] S. Hasipek, N.E. Senisik, E.S. Çetin, An Examination of Bacterial Colonisation on Nickel-Titanium Arch-wires with Different Surface Properties, J. Clin. Diagnostic Res. (2019), https://doi.org/10.7860/jcdr/2019/40377.12899.

[143] E. Petoumenou, M. Arndt, L. Keilig, S. Reimann, H. Hoederath, T. Eliades, A. Jäger, C. Bourauel, Nickel concentration in the saliva of patients with nickel-titanium orthodontic appliances, Am. J. Orthod. Dentofac. Orthop. 135 (2009) 59–65, https://doi.org/10.1016/j.ajodo.2006.12.018.

[144] M. Khaneh Masjedi, O. Niknam, N. Haghighat Jahromi, P. Javidi, V. Rakhshan, Effects of Fixed Orthodontic Treatment Using Conventional, Copper-Included, and Epoxy-Coated Nickel-Titanium Archwires on Salivary Nickel Levels: A Double-Blind Randomized Clinical Trial, Biol. Trace Elem. Res. 174 (2016) 27–31, https://doi.org/10.1007/s12011-016-0690-7.

[145] E. Petoumeno, M. Kislyuk, H. Hoederath, L. Keilig, C. Bourauel, A. Jäger, Korrosionsverhalten und Nickelabgabe von Nickel-Titan-Drähten im Verlauf der klinischen Anwendung, J. Orofac. Orthop. 69 (2008) 411–423, https://doi.org/10.1007/s00056-008-0808-4.

[146] I. Ben Naceur, A. Charfi, T. Bouraoui, khaled Elleuch, Finite element modeling of superelastic nickel-titanium orthodontic wires, J. Biomech. 47 (2014) 3630–3638, https://doi.org/10.1016/j.jbiomech.2014.10.007.

[147] M. Gannoun, M.L. Hellara, C. Bouby, T. Ben Zineb, T. Bouraoui, Numerical simulation of the force generated by a superelastic NiTi orthodontic archwire during tooth alignment phase: comparison between different constitutive models, Mater. Res. Express 5 (2018), https://doi.org/10.1088/2053-1591/aaba4d 045405.

[148] N. Ulff, C. Bouby, A. Lachiguer, T. Bouraoui, T. Ben Zineb, Modeling of Hydrogen Effects on the Thermomechanical Behavior of NiTi-Based Shape Memory Alloys, Shape Mem. Superelasticity. 5 (2019) 206–217, https://doi.org/10.1007/s40830-019-00224-7.

[149] W. Elkhal Letaief, A. Fathallah, T. Hassine, F. Gamaoun, Finite element analysis of hydrogen effects on superelastic NiTi shape memory alloys: Orthodontic application, J. Intell. Mater. Syst. Struct. 29 (16) (2018) 3188–3198.

[150] M.F. Razali, A.S. Mahmud, N. Mokhtar, J. Abdullah, Influence of sliding friction on leveling force of superelastic NiTi arch wire: A computational analysis, AIP Conf. Proc. 2017 (1891) 1–7, https://doi.org/10.1063/1.5005455.

[151] M.F. Razali, A.S. Mahmud, Computational study on the effect of contact friction towards deactivation force of superelastic NiTi arch wire in a bracket system, Mater. Res. Express 6 (8) (2019) 085709.

[152] D. Baghdadi, S. Reimann, L. Keilig, C. Reichert, A. Jäger, C. Bourauel, Biomechanical analysis of initial incisor crowding alignment in the periodontally reduced mandible using the finite element methodBiomechanische Analyse der initialen Nivellierung eines Frontengstandes im parodontal reduzierten Unterkiefer mittels der Finite-Elemente-Methode, J. Orofac. Orthop. 80 (4) (2019) 184–193.

[153] M. Mehrpouya, A. Gisario, H. Huang, A. Rahimzadeh, M. Elahinia, Numerical study for prediction of optimum operational parameters in laser welding of NiTi alloy, Opt. Laser Technol. 118 (2019) 159–169, https://doi.org/10.1016/j.optlastec.2019.05.010.

[154] A. Nazarahari, D. Canadinc, Prediction of the NiTi shape memory alloy composition with the best corrosion resistance for dental applications utilizing artificial intelligence, Mater. Chem. Phys. 258 (2021), https://doi.org/10.1016/j.matchemphys.2020.123974 123974.

[155] S. Asadi, T. Saeid, F. Habibi, Y. Kim, A. Valanezhad, N. Park, Phase identification and thermodynamic modeling of a quaternary system in dissimilar laser-welded NiTi/ASS, J. Mater. Res. Technol. 16 (2022) 25–38, https://doi.org/10.1016/j.jmrt.2021.11.154.

[156] R. Liu, X.i. Liu, J. Zhou, Q. Nie, J. Meng, J. Lin, S. Wang, Bioinspired Superhydrophobic Ni–Ti Archwires with Resistance to Bacterial Adhesion and Nickel Ion Release, Adv. Mater. Interfaces 6 (7) (2019) 1801569.

[157] K. Venkatesan, V. Kailasam, S. Padmanabhan, Evaluation of titanium dioxide coating on surface roughness of nickel-titanium archwires and its influence on Streptococcus mutans adhesion and enamel mineralization: A prospective clinical study, Am. J. Orthod. Dentofac. Orthop. 158 (2020) 199–208, https://doi.org/10.1016/j.ajodo.2019.07.019.

[158] C. Zhang, X. Sun, X. Hou, H. Li, D. Sun, The corrosion resistance of composite arch wire laser-welded by NiTi shape memory alloy and stainless steel wires with Cu interlayer in artificial saliva with protein, Int. J. Med. Sci. 10 (2013) 1068–1072, https://doi.org/10.7150/ijms.5878.