



ISSN: 0975-833X

Available online at <http://www.journalcra.com>

International Journal of Current Research
Vol. 10, Issue, 12, pp.76573-76579, December, 2018

DOI: <https://doi.org/10.24941/ijcr.33600.12.2018>

**INTERNATIONAL JOURNAL
OF CURRENT RESEARCH**

RESEARCH ARTICLE

ORTHODONTIC WIRES: A RECENT ANATOMIZATION

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ARTICLE INFO

Article History:

Received 18th September, 2018

Received in revised form

06th October, 2018

Accepted 25th November, 2018

Published online 31st December, 2018

ABSTRACT

Literature aims to have a comprehensive representation of wide array of orthodontic wires available for correcting dental irregularities or preserving the maintaining the existing tooth or teeth positions. The properties of these materials is explained in terms of stress strain relationship in response to external load. Wires considered to be one of the active components differ on the basis of various mechanical and physical properties under various stages of operation. The wires discussed are cobalt chromium wires, stainless steel wires, nickel titanium wires, alpha titanium wires and gold wires.

Key Words:

Orthodontic Wires, Stress – Strain, Stainless Steel, Nickel-Titanium, Cobalt-Chromium, Gold Wires.

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Citation: Dheerendra Kumar Singh, Pavan Kumar Singh, Komal Khatri, Priyashree Joshi, Soumendu Bikash Maiti, Pragnajyoti Das, 2018. "Orthodontic wires: a recent anatomization", *International Journal of Current Research*, 10, (12), 76573-76579.

INTRODUCTION

Metals are amazing considering their ability to be rolled into sheets as thick as the hulls of ships or as thin as gold and aluminum foil, to be drawn into wire cables supporting bridges or into fine strands, one-half the thickness of a human hair, for delicate electronic instruments; to be softened by heat and hardened by cold working. Metals are cast into dental inlays and molded into grant steam shovels. They resist wear and corrosion; they conduct heat and electricity; they may be joined or machined; they have luster, yet are generally inexpensive. Finally, metal can be shaped into the myriad of attachments, archwires, and instruments that comprise the entire armamentarium, which permits the technical procedure of orthodontics (Brantley, 1997).

Criteria of an ideal arch wire (Cramb, 2002; Humphrey, 1935): Several characteristics of orthodontic wires are considered desirable for optimum performance during treatment. These include

- High strength
- Low stiffness
- High range
- High spring back
- High formability
- High stored energy
- Biocompatibility
- Environmental stability
- Low surface friction
- Poor biostability
- Resilience
- Weldable and solderable

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- Esthetics
- Economical
- Corrosion resistance

The basic properties of elastic materials is defined in terms of its stress-strain response to an external load. Both stress and strain refer to the internal state of the material being studied. The various properties of materials are elastic limit, proportional limit, yield strength, tensile strength, modulus of elasticity, load deflection rate, stiffness, springback, resilience, formability, ductility, biocompatibility, twinning, hysteresis, quenching and annealing. Crystals or grains of metals and alloys are composed of billions upon billions of atoms regularly arranged in a space lattice. Crystals of different metals vary in size from those that can be seen with the naked eye to those that require high powered magnification (Anusavice, 1996; Kapila, 1989).

Table 1. The fourteen space (Bravais) lattices

Crystal System	Space Lattice
Cubic	Simple cubic
	Body-centered cubic
	Face-centered cubic
Hexagonal	Simple hexagonal
Monoclinic	Simple monoclinic
	Base-centered monoclinic
Orthorhombic	Simple orthorhombic
	Body-centered orthorhombic
	Face-centered orthorhombic
	Base-centered orthorhombic
Rhombohedral (Trigonal)	Simple rhombohedral
Tetragonal	Simple tetragonal
	Body-centered tetragonal
Triclinic	Simple triclinic

Cobalt chromium alloys: After gold alloy archwires were largely supplanted by stainless steel archwires in the early 1960s, there was a period in which only one other alloy competed for a market share- cobalt-chromium a (Co Cr) alloys. Such alloys belong to a group called Stellites. Stellite alloys are renowned for a corrosion resistance that exceeds that of stainless steel and is at least comparable to that of the venerable gold alloys (Elgiloy Company, 1977). In wrought alloys, such as those used to make archwires (different from cast alloys used in prosthodontics) 40%-45% cobalt and 15%-22% chromium are typical; other elements may also be present, including nickel for strength and ductility and iron, molybdenum, tungsten and titanium to form stable carbides that enhance hardenability and set resistance. One particular alloy, Elgiloy, was patented by the Elgin National Company (Lincoln, Neb) as the main spring of their analog watches with "the heart that never breaks". That particular alloy contained 8 elements, the most important of which were Co (40%), Cr (20%), Ni (15%), and Fe (16%).⁴² At some point scientists at Rocky Mountain Orthodontics perceived that the use of elgiloy for springs, drive bands, torsional bars, ball bearings, and cables could be valuable to orthodontic practitioner, since archwires were routinely used in some of these modalities during patient treatment.

Composition:

Cobalt 40%
chromium 20%
nickel 15%

molybdenum 7%
manganese 2%
carbon 15%
beryllium 0.4%
iron 15.8%

In addition to having similar stiffness characteristics as stainless steel, the alloy was capable of having its strength, and more importantly its formability, modified by heat treatment. At that time practitioners particularly welcomed the formability before heat treatment in order to bend loops into the wires and to enhance the working ranges of their otherwise somewhat rigid appliances. Once the appliance was fabricated, however, the practitioners no longer required the formability. Instead they desired resilience in order to capitalize on the inherent elasticity of the material, which could be achieved by heat treating the alloy at 482 °C for 7- 12 min. This so called precipitation hardening heat treatment increased the ultimate strength and resilience of these archwires without changing the stiffness. Indeed four such products were eventually marketed that varied in initial formability from a hard and resilient high spring temper alloy, designated "red," to a soft and formable alloy, designated "blue." Two of these, the blue alloy and a slightly less formable but nonetheless ductile "yellow" alloy were developed between 1958 and 1961 in order to match, in their heat treated states, the temper of the standard and extra hard stainless steel of the day. At that time in the evolution of the materials, wire shape and size dominated over structural properties, owing largely to the fact that the stiffness of all the materials were virtually identical (Elgiloy Company, 1977; Filmore, 1976; Filmore, 1979).

Types of Elgiloy: Cobalt chromium alloys are available commercially as Elgiloy (RMO), Azura (ORMCO), and Multiphase (American Orthodontics). Elgiloy is manufactured in 4 tempers: Soft (blue), ductile (yellow), semi resilient (green), and resilient (red) in increasing order of resilience.

Blue Elgiloy: Softest of the four wire tempers. Can be bent easily with fingers or pliers. Recommended for use when considerable bending, soldering or welding is required.

Yellow Elgiloy: Relatively ductile and more resilient than blue elgiloy. Can also be bent with relative ease. Further increase in resilience and spring performance can be achieved by heat treatment.

Green Elgiloy: More resilient than yellow elgiloy. Can be shaped with pliers before heat treatment

Red Elgiloy: Most resilient and provides high spring qualities. Careful manipulation with pliers is recommended when using this wire because it withstands only minimal working. Elgiloy had 2 potential advantages over its stainless steel competitor: it could be delivered to the practitioners in different degrees of hardening, or temper, and it could be further hardened by heat treatment. The stiffness of each temper should be identical for the same wire configuration and sizes, but the strength should increase as the tempers proceeded from the "blue" temper level to the "yellow," "green," "red" levels. Additionally the plasticity should vary as the names would suggest namely, soft (blue), ductile (yellow), semi resilient (green), and resilient or hard (red).

Thus the practitioner has the option of selecting the amount of formability that was required during placement (Elgiloy Company, 1977; Thurow, 1972).

Mechanical Properties: Mechanical properties of Co-Cr wires are very similar to those of stainless steel arch wires (Kapila, 1989). Spring back – with exception of red temper elgiloy, non heat-treated Co-Cr wires have a smaller springback than stainless steel wires of comparable sizes. But this property can be improved by adequate heat treatment. Stiffness: High Module of elasticity of Co-Cr and stainless steel arch wires suggest that these wires deliver twice the force of Beta titanium and 4 times the force of Nitinol wires for equal amounts of activation.

Formability: Co-Cr wires have good formability that can be modified by heat treatment. During that time practitioners appreciated the formability before heat treatment in order to bend loops into wires. However, once appliance was fabricated, the practitioner no longer required formability. Instead they required resilience in order to capitalize on inherent elasticity of material which could be achieved by heat treatment.

Joinability: Can be soldered and welded. Caution should be exercised when soldering attachments to these wires since high temperature (749°C) causes annealing with resultant loss in yield and tensile strengths. Low fusing solder is recommended. Biocompatibility and Environmental Stability Good.

Friction: Although longer frictional forces have been noted previously between brackets and Co-Cr wires but recent reports suggest that resistance to tooth movement along stainless steel and Co-Cr wires may be comparable.

Recovery Heat Treatment: Once wire bending was done, formability was no longer needed if the wire was to effectively move teeth. Instead resilience was sought from the same wire. This was available in Elgiloy if, after plastic deformation, the wire was heat treated in an oven for 5 hours for 480 degree C.

Ordinarily the wires are heat treated before being supplied to the user and ordered in several degrees of hardness (soft, ductile, semi spring temper, spring temper). The orthodontist can, in addition, heat-treat the wires either in an oven or by passing an electric current through the wire by using certain type of spot welders. A typical cycle would be 482°C (900°F) for 7-12 minutes. This heat treatment would increase the yield strength and decrease the ductility. Heat treatment at temperatures above 749°C(1200°F) results in a rapid decline in resistance to deformation because of partial annealing. Optimum levels of heat treatment are confirmed by a dark straw colored wire or by the use of temperature-indicating paste.

Clinical Application: Elgiloy is used as an orthodontic wire because it is easier to bend than stainless steel, NiTi and Beta Titanium in its “as received state”. For this reason it is often preferred in techniques in which loops in archwires are used. Can be heat treated after manipulation to achieve a hardness approximately equal to that of stainless steel. Along with stainless steel it is considered most ideal and economical finishing wire. The advantages over stainless steel wires are greater resistance to fatigue and distortion and longer function

as a resilient spring. In other aspects mechanical properties of Co-Cr wires are very similar to those of stainless steel arch wires. Therefore SS wires may be used instead of Co-Cr wires of same size in clinical situations in which heat hardening and added torsional strength of Co-Cr wires are not required. The drawback is however high elastic force delivery, similar to that of stainless steel (Elgiloy Company, 1977; Filmore, 1976; Filmore, 1979)

Stainless Steel Archwires: Stainless steel entered dentistry in 1919, being introduced at Krupp’s Dental Polyclinic in Germany by the company’s dentist Dr. F. Hauptmeyer. He first used it to make a prosthesis and called it Wipla (Wieplatin; in German like platinum), the designation under which it is still used in Europe. By the 1960s, gold was universally abandoned in favor of stainless steel. This is how stainless steel was marketed in lieu of gold: (1) the force per unit activation of stainless steel was greater than that of gold (i.e., high stiffness was an advantage they claimed); and (2) by being smaller in size, stainless steel appliances were regarded as being more esthetic than gold appliances. Stainless steel also had excellent corrosion resistance, work-hardening capabilities, and a frictional magnitude that was so low that it became the standard of the profession. Stainless steels constitute the major alloy system used in orthodontics. However, the metallurgy and terminology of these alloys are intimately connected to that of the simpler binary iron-carbon alloy system and the ordinary carbon steel alloys (Zapffe, 1949; Kohl, 1964; ASM, 1961; Craig, 1977). In addition to iron and carbon the stainless steels contain chromium (at least 10-13%) which improves corrosion resistance. This is achieved by the passivating effect in which the chromium exposed at the surface of the alloy is readily oxidized to form a tenacious surface film of chromic oxide. This film resists further attack from aqueous media thus preventing corrosion. When at least 8% Nickel was present, the single-phase structure of austenite was stabilized, and the overall corrosion resistance was enhanced⁹. Carbon content was purposely maintained below 0.20% to reduce the formation of chromium carbides, structures that can ultimately foster the corrosion of austenitic steels. The addition of Cr and Ni to steel causes the critical temperature (T_c) to be lowered. If sufficient quantities of these two metals are incorporated, the austenitic structures remains even at room temperature.

One of the most commonly used SS contains 18% Cr and 8% Ni (18/8 SS). This alloy has a critical temperature below the point at which atomic movements are possible and is therefore referred to as austenitic stainless steel. These applications involve a degree of cold working since bending, drawing or swaging shapes the alloy. This result in the formation of a wrought structure. It is this property coupled with a relatively high modulus of elasticity that makes wrought structure wires suitable for orthodontics (Craig, 1977; William, 1977; Denny, 1993). Research using X-ray diffraction has shown that austenitic stainless steel orthodontic wires may not always possess the single-phase austenitic structure that is based upon a face-centered cubic (fcc) arrangement of the iron atoms.^{1,16} Stainless steel requires a higher temperature for annealing (1800° to 2000° F or 1000° to 1100° C) than does carbon steel. A most important heat treatment process for orthodontic stainless steel is the relatively low temperature process of stress relieving, which is used both in manufacturing and at the time of clinical application. Prevention of intergranular corrosion is one of the most important problems in

manipulating stainless steel at high temperature.⁸ Two means of prevention are available: Keeping out of the sensitizing temperature range (800° to 1200° F or 425° to 650°C).

Controlling the carbon content: Stabilized steel is less susceptible to intergranular corrosion, but it is still not 100% safe. Improper handling by the user can easily nullify the advantage. These steels are stabilized as they are delivered, but there is no compensation for the carbon introduced later. In orthodontics a plentiful supply of carbon is always available in the organic debris present in the mouth. Leaving these deposits on the piece of steel as it is welded or soldered will immediately carbonize them and infuse the surface of the steel with carbon. Extreme care should be taken to avoid this and any other carbon contamination of stainless steel that is to be heated. Three grades of stainless steel wires are usually required. Soft, half-hard and hard (Anusavice, 1996; Funk, 1951) Australian Arch Wires (Wilcock, 1989; Wilcock, 1988). After many years of research and development in producing high tensile wires, Mr. Wilcock finally produced a Cold drawn heat treated wire that combined the balance between hardness and resiliency with unique property of Zero stress relaxation that Dr. Begg was seeking. Different grades of Australian wires formerly available were:

- Regular
- Regular plus
- Special
- Special plus

Until recently grade of wire routinely used was special plus and for those cases resistant to bite opening Extra Special Plus (ESP) was used. Recently A.J. Wilcock Scientific & Engineering Co., the manufacturers of this wire have announced new series of wire grades and sizes. The fundamental difference for the superior properties for these new wires is use of new manufacturing process called pulse straightening as against spinner straightening. A.J. Wilcock Jr. states that wires are straightened by use of two processes. Spinner straightening. Pulse straightening process

Gold wires: Pure Gold is the noblest of all dental metals, rarely tarnishing and corroding in the oral cavity. It is inactive chemically, and it is not affected by air, heat, moisture and most solvents. It is the most ductile of all metals, as demonstrated by its ability for a 1oz cylinder to be drawn into a wire 100 km long in length. It is the most malleable of all metals, as shown by its ability to be hammered to a thickness of 0.00013 mm, about one third of thinnest gold foil used in dentistry. There are two types of Gold wires recognized in American Dental Association (ADA) specification no 7, year 1984.

Type I: They must contain at least 75% gold and platinum group metals.

Type II: They must contain at least 65% gold and platinum group metals.

In addition to Type I and II Gold wires used in orthodontics before 1950, two other types of wires were also used with high content of Gold in at least one of them.

Palladium-Gold-Platinum (P-G-P): Because of their high fusion temperature and therefore high crystallization

temperature, they are especially useful as wires to be cast against and meet the composition requirements for an ADA type I wire.

Palladium-Silver-Copper (P-S-C): These wires are neither Type I nor Type II gold wires, but their mechanical properties would meet the requirements for an ADA Type I or Type II alloy. The corrosion resistance of palladium-silver dental alloy, both in cast and wrought forms, is generally satisfactory.

Fusion Temperature: According to ADA specification no 7, for a type I wire, this temperature is 955⁰ C (1751⁰ F) or higher, for the type II wire the minimum fusion temperature should be 871⁰ C (1600⁰ F). The modulus of elasticity of wrought gold wires is in the range of 97,000 to 117,000 Mpa (14,000,000 to 17,000,000 Psi) which is slightly higher than that for gold casting alloys. It increases by approximately 5% after a hardening heat treatment. All gold alloy wires that contain copper are heat treatable as the Gold casting alloys. Type I and II alloys usually do not harden, or they harden to a lesser degree than do the type III and IV alloys. Gold alloy is placed in an electric furnace for 10 min at a temperature of 700⁰ C or 1292⁰ F. This is called as annealing. Then it is quenched in water. During this period all intermediate phases are presumably changed to a disordered solid solution, and the rapid quenching prevents ordering from occurring during cooling. The tensile strength, proportional limit and hardness are reduced by such a treatment but the ductility is increased. The age hardening or hardening heat treatment of dental alloys can be accomplished in several ways. One of the most practical hardening treatments is by “soaking” or ageing the alloy at a specific temperature for definite time, usually 15-30 minutes, before it is water quenched. The ageing temperature depends upon the alloy composition but is generally between 200⁰ C (400⁰ F) to 450⁰ C (840⁰ F). The proper time and temperature are specified by the manufacture. Cold working is also the usual method of hardening gold alloy. Much more cold working is required for Gold alloys than Steel to harden it. This is to adjust the drawing and annealing schedule to compensate (Brantley, 1997; Anusavice, 1996).

Multistranded stainless steel wires: Multistranded wires are made of stainless steel and composed of specified numbers of thin wire sections coiled around each other to provide a round or rectangular cross-section. The multistranded wire was also 25% stronger than the 0.010-inch stainless steel wire. The 0.0175-inch triple-stranded wire and 0.016-inch nitinol wire demonstrated similar stiffnesses. However, nitinol tolerated more than 50% greater activation than the multistranded wire. The triple-stranded wire was also half as stiff as a 0.016-inch beta-titanium wire. In a more recent investigation, Kusy and Stevens state that although the elastic properties of multistranded wires vary widely, several of these wires compare favorably with some of the beta-titanium and nitinol wires. Compared to conventional NiTi wires, current multistranded SS products generally matched the stiffnesses but had lower strengths and ranges, both of which are proportional to O_{PL} . For multistranded SS wires, a theoretical investigation assumed values for O_{PL} from 1.03 to 3.28 GPa, although the measured O_{YS} values here averaged 1.83 GPa for the triple wires and 1.78 GPa for the coax wires. The coax geometry was introduced to provide lower stiffness and higher range, but the physical properties of currently used SS alloys result in no improvement in range over single SS leveling wires.

For conventional NiTi wires, the same theoretical investigation assumed a value of $O_{PL} = 1.24$ GPa, which is within 15% of the measured O_{EL} average of 1.10 GPa. These NiTi wires had 2-3 times the strength and range of the multistranded SS wires ironically, the present O_{EL} values average 2-3 times higher than previously reported values. Multistranded stainless steel (SS) archwires do not match the strength and range of conventional NiTi wires because they are fabricated with SS alloys that possess moderately high yield strengths (Funk, 1951; Wilcock, 1989; Wilcock, 1988; Kusy, 1984). When comparing stainless steel and nickel-titanium of similar size or of similar number of strands in laboratory tests (within-category comparisons), the results show, in general, considerably greater stiffness for stainless steel-about 4 to 5 times higher than that of superelastic NiTi. The difference in performance between a NiTi alloy and a multistranded stainless steel is detectable mainly with the increase of loading (Begg, ?; Parul Taneja et al., 2003).

Alpha titanium wires: These wires were developed by A.J. Wilcock Jr. in 1988 and claimed to have extraordinary resilience whilst maintaining formability. Like the stainless steel and nickel-titanium orthodontic wires pure titanium has different crystallographic forms at high and low temperatures. At temperatures below 885°C, the stable form is α -titanium, which has the hexagonal closed pack crystal structure, whereas at higher temperatures the stable form is β -titanium, which has the bcc structure. The elastic modulus and yield strength at room temperature for α -titanium is approximately 110 GPa and 40 MPa respectively. Certain elements, such as aluminum, carbon, oxygen and nitrogen, stabilize the α -titanium structure. That is, they raise the temperature for transformation to β -titanium. The composition of α -titanium include 88.9% titanium, 7.86% Aluminum and 4.05% Vanadium. Hexagonal lattice possesses fewer slip planes making it less ductile than β -titanium. Alloy is strictly near alpha phase of titanium rather than pure alpha titanium alloy because there is certain amount of beta phase retained in them at room temperature. The wires are soft enough for initial gentle action on teeth in spite of large wire dimension as also for intraoral activation. They seem to harden and become brittle with passage of time in the mouth, possibly due to the absorption of hydrogen and formation of titanium hydrides (Wilcock, 1989).

Beta – titanium wires: Like Stainless steel and Nitinol, pure titanium has different crystallographic forms at high and low temperature. At temperatures below 885°C [1625°F], the hexagonal close packed (HCP) or α lattice is stable, whereas at higher temperatures, the metal rearranges into a body centered cubic (BCC) or β crystal. The beta titanium alloys exhibit a bcc allotropic form of titanium (called β) that is metastable at room temperature due to the presence of alloying elements such as molybdenum, vanadium, columbium, tantalum, manganese, tin, iron, chromium, cobalt, Nickel and Copper. In 1979, Goldberg and Burstone presented their preliminary work investigating the feasibility of using beta-stabilized titanium alloys for orthodontic applications. After much testing, Goldberg and Burstone determined that, with proper processing, 11% molybdenum, 6% zirconium, and 4% tin beta titanium alloy would be usable in orthodontics. The modulus of elasticity of their wire, later called TMA (Titanium Molybdenum Alloy) was 9.4×10^6 P.S.I. or 41% that of stainless steel.¹⁴ It could be bent 105% further or twice the distance of stainless steel before reaching its yield point.

Beta titanium cannot be soldered, but it can be welded to itself with minimal strength change. The high ductility [ability of the material to withstand permanent deformation without rupture under tensile load] of β -Titanium allow it to be formed into arches with tieback loops or segments with complicated loop configuration. It also offers the possibility of varying force magnitude by a choice of material rather than cross section of the wire. Beta Titanium is unique in comparison to commonly used orthodontic wire in that it allows direct welding of auxiliaries to an arch wire without reinforcement by soldering. The range for optimal voltage setting for different TMA wire combination 9.5 to 8 (Graber, 2000; ASM Committee on Wrought Stainless Steels, 1961; Burstone, 1981; Bagden, 1997).

Nickel titanium archwires: The engineering of nickel-titanium (NiTi) alloys has made remarkable progress since the original work of Buehler for the Naval Ordnance Laboratory in the early 1960s. Buehler's preliminary results led to development of the first NiTi orthodontic alloy 55% nickel and 45% titanium by pioneers such as Andreasen and his colleagues.

- Composition:
- Nickel-54%
- Titanium-44%
- Cobalt-2%

When stainless steel was the predominant material used for fabrication of orthodontic wires, control of forces generated by the wires was limited to the use of either varying wire cross-sections or complicated loop configurations to affect the stiffness characteristics of the appliance. This treatment strategy has been aptly termed "*variable cross-sectional*" approach to orthodontic treatment. Most recently, Ni free, Titanium – Niobium wires have been introduced as a finishing wire (Buehler et al., 1963; Buehler, 1968; Andreasen, 1972).

Classification of niti compounds (waters, 1992):

Based on Transformation Temperature Ranges

Group 1: Alloys with TTRs between room temperature and body temperature [Active Martensite].

Group 2: Alloys with TTR below room temperature [Austenite active]

Group 3: Alloys with TTR close to body temperature, "which by virtue of the shape memory effect spring back to their original shape when activated by body heat". Based on chronologic phases with a specific mode of force delivery (Evans, 1996).

Phase 1: Gold and Stainless steel Alloys

Phase 2: Beta Titanium and stabilised NiTi [Nitinol]

Phase 3: Active Austenite [Superelastic NiTi alloys]. [Predominantly Austenitic at room temperature].

Phase 4: Thermodynamic – active martensitic, and

Phase 5: Graded thermodynamic

The complex metallurgical nature of the nickel-titanium orthodontic wires and its relevance to clinical applications has spawned many research investigations.⁵ The nickel-titanium wires contain approximately equiatomic proportions of nickel and titanium, and are based upon the intermetallic compound NiTi (sometimes written as TiNi). There are two major NiTi phases in the nickel-titanium wires. Austenitic NiTi (hereafter termed austenite) has an ordered bcc (CsCl-type) structure (sometimes termed B2) that occurs at high temperatures and low stresses. Martensitic NiTi (hereafter termed martensite) has been reported to have a distorted monoclinic, triclinic, or hexagonal structure, and forms at low temperatures and high stresses. The shape-memory effect (SME) is associated with a reversible martensite-austenite transformation, which occurs rapidly by crystallographic twinning at the atomic level. The transformation from austenitic to the martensitic phase (thermoelastic martensitic transformation) is reversible, unlike in metals. In a shape memory alloy, transformation from an austenite to a martensite phase can also occur by application of stress within a defined temperature range. The martensite so-formed is called stress-induced martensite (SIM), and the driving force for the transformation is mechanical, as opposed to thermal. Within a temperature range, martensite can be made stable with the application of stress, but it becomes unstable again when the stress is removed. There is a linear relationship between stress and temperature with respect to induction of martensite; a decrease in temperature is equivalent to an increase in stress. There is a difference in the transformation temperatures upon heating from martensite to austenite and cooling from austenite to martensite, resulting in a delay or "lag" in the transformation.

This difference known as the transformation temperature hysteresis, is generally defined as the difference between the temperatures at which the material is 50% transformed to austenite upon heating and 50% transformed to martensite upon cooling. For NiTi Alloys, the difference between M_p and A_p is 25-50°C. Thus Nitinol transformations exhibit thermal hysteresis, $M_s \neq A_f$ and $M_f \neq A_s$. The use of rectangular NiTi wires has been proposed to obtain early torque control during the aligning phase of treatment. However, in terms of sensitivity to interbracket distance, torsional stress is considerably different from flexural stress. In flexural stress, modifications in the length of the specimen generates linear changes in the forces. As a consequence, the results available from experiments studying flexural behaviors cannot be indiscriminately applied to the torsional behavior of NiTi wires (Graber, 2006).

The original Nitinol, developed in the early 1960s, is a stabilized form of the alloy in which work hardening has abolished the phase transformation. The alloy does not exhibit a shape memory effect, but the low modulus of elasticity and the high working range make Nitinol useful when considerable deflections are necessary. This alloy was passive, as the SME had been suppressed by cold working the wire during drawing to more than 8 to 10%⁷ what was so attractive about this martensitic stabilized alloy was its low force per unit of deactivation – that is, its low stiffness compared with other alloys, this wire was quite springy – delivering only 1/5 – 1/6 the force per unit of deactivation and thereby better meeting the criterion of light, continuous force. The wire has moderately high yield strength of 250000 psi with a low modulus of elasticity 4800000-psi. These values give Nitinol a good working range or spring back (Proffit, 2001).

A major limitation was its lack of formability. The poor formability of these wires implies that they are best suited for pre-adjusted systems. Although nitinol feels quite flexible and gives the impression that it is very ductile, it will readily break when bend over a sharp edge (Brantley, 1997).

Active Austenitic Nickel Titanium Alloys:

Chinese NiTi

Japanese NiTi [Sentalloy]

Copper NiTi 27°C

CONCLUSION

Recent advances in orthodontic arch wires proves a clear commitment to high performance standard, lifelong learning and strict accreditation in search of an ideal wire. To obtain benefit of optimum and predictable treatment results one can depend on selection of appropriate wire size and alloy type. Fiber reinforced composites are regarded as the last great frontier of orthodontic materials. Due to their excellent aesthetics and strength, as well as the ability to customize their properties to the needs of the orthodontist, they are expected to replace metals in orthodontics, just as composites have replaced aluminium in the aircraft industry. No doubt, these materials promise several exciting new possibilities in biomechanics, and could revolutionize the practice of orthodontics.

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