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Review of Superelastic Differential Force Archwires for Producing Ideal Orthodontic Forces: an Advanced Technology Potentially Applicable to Orthognathic Surgery and Orthopedics

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Abstract

Purpose of Review Gentle and continuous loads are preferred for optimum orthodontic tooth movement. Nitinol, an alloy of nickel and titanium developed for the aerospace industry, found its first clinical applications in orthodontics because it has ideal load-deflection behavior. The purpose of this review is to elucidate the criteria for effective orthodontic mechanics relative to emerging Nitinol technology. The specialized materials with variable stiffness that were originally developed for orthodontics are increasingly attractive for in the temporomandibular joint, orthognathic surgery, and orthopedics.

Recent Findings The evolution of orthodontic archwires is driven by a need to achieve low load-deflection characteristics and Nitinol is the alloy of choice. Scientific knowledge of the biological response to orthodontic forces continues to grow, but definitive guidance on optimal force levels for individual teeth is elusive. Finite element models (FEM) that take into account periodontal ligament (PDL) stresses indicate differential force archwires are needed to realize optimal treatment. However, previous wire fabrication methods, including welding of different materials and selective resistive heating, are limited by poor mechanical performance and spatial resolution. Recently, a novel laser processing technique was developed for precisely programing relative levels of stiffness in a single archwire. FEM was used to estimate the optimal force for each tooth by calculating the 3D bone-PDL surface area.

Summary There remains a general consensus that light and continuous forces are desirable for orthodontic treatment. New developments in archwire materials and technology have provided the orthodontist with a complete spectrum of load-deflection rates and differential force options to express these forces with maximized archwire economy. These technologies also appear to have application to orthopedic implant devices.

Keywords Orthodontic archwires · Superelastic · Nitinol · Shape memory alloy · Laser processing

Introduction

The orthodontist is faced with a dizzying array of archwire choices relative to materials, shapes, and sizes. Since Angle's introduction of the edgewise appliance in 1928, research and commercial efforts have focussed on understanding the biomechanics of tooth movement, specifically the interactions between the orthodontic archwire and the biological tissue supporting the rigid tooth. Basically, the orthodontic appliance sets up a force system applied to the tooth which in turn induces compressive and tensile stresses and strains in the supporting tissues. Thus, the selection of archwire material, shape, and size to provide biologically appropriate force to the tooth is an important aspect of developing an orthodontic treatment protocol. Recent advances in materials science have provided even more choices than ever before. Orthodontists are tasked with understanding the mechanical response of these advanced materials in the same way a civil engineer would select materials for building a structure.

The purpose of this review is to identify recent developments in materials for expressing orthodontic forces through archwires and related technology. A brief overview is given of important historic contributions in the development of materials that provide desirable mechanical properties, including

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low stiffness and a high elastic strain limit. This review will also examine recent contributions, relative to the classic articles that have defined the discipline, to focus on ideal force ranges for efficient tooth movement throughout the arch. Finally, opportunities for additional research and clinical practice of these specialized materials in a variety of medical and surgical applications well beyond orthodontics will be explored.

Archwires for Continuous Forces

In orthodontic treatment, it has long been held that continuous gentle forces are desired for optimum tooth movement [1–5]. When applied force is extremely low and/or intermittent, there is insufficient stress in the periodontal ligature (PDL) to trigger tooth movement. Conversely, when forces are too high, excessive compressive stress in the PDL causes hyalinization and tissue necrosis, thereby blocking tooth movement. It was proposed that there is an optimal range of forces applied to the tooth over which the stresses in the PDL support efficient tooth movement through a process historically defined as "bone remodeling" [1]. In a more modern perspective, tooth movement actually involves three separate mechanically induced osseous effects: anabolic modeling (formation), catabolic modeling (resorption), and remodeling (turnover). The rate-limiting step in tooth movement is bone resorption at the PDL interface in the path of the applied load [2]. The merit of an optimum force range will be discussed later in this article, but to understand the context of modern archwire development, we will assume that it is true for the time being. Then, to minimize the costs of repetitive archwire adjustments and changes and maximize efficiency of tooth movement, it is desirable to utilize materials that deliver forces within this optimal range over the entire activation distance [5]. Hooke's law describes force that is proportional to strain while Burstone [5] defined the *load-deflection rate* as the amount of orthodontic force drop-off experienced with deactivation of the wire during tooth movement. Thus, an ideal archwire delivers force with a low load-deflection rate that has a large range of action.

Because the load-deflection rate is related to the mechanical properties of the archwire material, new alloys have been in continuous development since the inception of modern edgewise orthodontics. Stainless steel replaced the noble metals due to lower cost, corrosion resistance, and good ductility; however, this substitution came at a cost of relatively high load-deflection rate. Frequent wire changes are required to re-energize the wire after short tooth movements. To improve this shortcoming, cobalt-chromium-based super alloys with much lower load-deflection rate were adapted for orthodontic use as an alternative to stainless steel (i.e. Elgiloy®, Rocky Mountain Orthodontics, Denver CO) [6]. Both stainless steel and cobalt-chromium archwires typically require the orthodontist to add a series of time-consuming bends before ligating an archwire into the brackets. These two materials are still in common use today; however, two developments in the 1970s significantly affected the mix of materials in orthodontic use today.

Nitinol Archwires

In 1961, the discovery of Nitinol was a precursor to a landmark change in orthodontic materials. Buehler et al. [7] observed the shape memory effect in near-equiatomic nickel-titanium alloys, creating significant interest in its use for biomechanical applications. Nitinol is a biocompatible alloy, suitable for medical devices exhibiting a temperature-dependent pseudoelastic (superelastic) property [8]. The two stable phases of Nitinol are martensite (low temperature) and austenite (high temperature), which are separated by a transformation temperature range (TTR). In the martensitic phase, the wires are soft and pliant; however, when heated through the TTR into the austenitic phase, the wires exhibit high strain recovery and resilience. Austenite also undergoes a stress-induced martensitic transformation, resulting in a pseudoelastic stress plateau in both loading and unloading up to strains of 10%. This pseudoelasticity compares remarkably well with biological materials, such as the bone and tendon, and has a theoretically ideal load-deflection rate [9].

In 1971, the first application of Nitinol in a medical device was an orthodontic archwire [10]. Andreasen and Barrett [11] used compliant Nitinol wires, which, due to fully cold-worked mechanical treatment, did not exhibit pseudoelasticity; nevertheless, the load-deflection rate was much lower than for stainless steel or cobalt-chromium wires. Furthermore, they noted the far greater activation range for Nitinol wires and postulated that between two and four stainless steel wires could be replaced during initial alignment. In further work, Andreasen and Morrow [12] pointed out that larger size Nitinol wires could be used earlier in treatment, better filling bracket slots for improved control in tooth leveling and rotation, all without causing increased patient discomfort. One significant drawback of the Nitinol wires is their limited formability, preventing the clinician from making detailed bends for controlled tooth movement [13], and under some physiologic conditions, they were also susceptible to fatigue fracture [14]. Another development during the 1970s, however, would change the course of treatment protocol and open the door for widespread Nitinol adoption [15].

Nitinol is not formable like stainless steel and cobalt-chromium; however, Andrew's [15] work on the six keys to normal occlusion leads the way to the so-called straight-wire treatment philosophy where a fully programmed bracket base with varying inclination angles replaced wire bending.

Growing adoption of various prescriptions of pre-programmed bracket systems reduced the need for detailed wire bending in the first and second phases of orthodontic treatment, enabling adoption of Nitinol as the wire of choice for straight-wire procedures. This along with continued improvements in material properties furthered the adoption of Nitinol in orthodontics.

The potential of Nitinol archwires to deliver pseudoelastic behavior was not realized until advances in material processing lead to the introduction of so-called *Chinese NiTi* by Burstone and Morton [16] in 1985 and *Japanese NiTi* by Miura et al. [17] in 1986. These modern Nitinol alloys had finely tuned thermomechanical processing, enabling the desirable pseudoelastic response (superelasticity). Shifting the TTR through heat treatment has been shown to vary the magnitude of the unloading stress providing a wide range of continuous force levels in wires with similar cross sections [18]. Additions of small amounts of copper have been substituted for nickel in the chemical composition of Nitinol. The copper substitution further reduces the slope of the unloading phase: pseudoelastic stress plateau. There is a lower stress hysteresis, without significantly affecting the transformation temperature of the material [19]. Copper NiTi archwires offer the desired combination of an ideal load-deflection rate with a long activation range.

Beta-Titanium Archwires

In 1979, Goldberg and Burstone [20] introduced a beta-titanium alloy archwire that, while stiffer than Nitinol, had a lower load-deflection rate than stainless steel and cobalt-chromium, and good formability characteristics. The beta-titanium alloy (Ti-11.3Mo-6.6Zr-4.3Sn) was amenable to complex bends and exhibited desirable properties for the finishing phase of orthodontic treatment [21]. These titanium-molybdenum-based beta-titanium alloys have been widely used in orthodontic practice. Sensitivity to the nickel found in stainless steel and Nitinol was a driving force for continued development of nickel-free archwires. Variations in beta-titanium compositions have been evaluated in several recent studies. Titanium-niobium alloys were found to exhibit similar mechanical properties to titanium-molybdenum by Dalstra et al. [22]. Suzuki et al. [23] found that a titanium-molybdenum-aluminum alloy wire produced a lower load-deflection rate than cold-worked Nitinol wires. Clinical trials in a rat population did not show any significant difference in tooth movement rates.

Recently, a new class of beta-titanium alloys has been introduced. These alloys, known as *Gum Metal* with a composition of Ti-36Nb-2Ta-3Zr-0.3O, exhibit a very low elastic stiffness [24]. Chang and Tseng [25] described the characteristics of Gum Metal archwires applied to orthodontic treatment, finding that the material expresses a super-low load-deflection rate and classifying it as superelastic. A comprehensive comparison of Gum Metal with titanium-molybdenum, pseudoelastic copper NiTi, and cold-worked Nitinol wires shows that, at low deflections, the load-deflection rate of Gum Metal is the lowest of any material [26]. Unfortunately, the study does not compare the load-deflection rate at higher strains where copper NiTi would express pseudoelastic properties and the load-deflection rate would drop to zero. The activation range of Nitinol alloys exceeds that of Gum Metal alloys; however, the ability to bend Gum Metal archwires with complex details provides an added advantage [27]. A double-blind randomized clinical trial did not find any significant difference in tooth movement between the Gum Metal and Nitinol archwires [28].

Contemporary View of Optimized Forces

Although the long-standing view that an ideal orthodontic force range produces optimized tooth movement is still quite popular in practice, recent publications have not been able to confirm or refute this theory. At least four competing hypotheses have been presented for the relationship between orthodontic forces and the rate of tooth movement [29]. In general, it is still widely debated whether a low force threshold exists for initiating tooth movement, whether tooth movement is proportional to applied force within any range of possible forces, and whether a high force threshold exists which would slow the rate of tooth movement. For instance, a split arch study in humans over 7 weeks of treatment compared pseudoelastic Nitinol to stainless steel archwires. A fourfold increase in applied force resulted in increased tooth movement over the study period without a measured increase in undesirable root resorption [30]. A comparable study in rabbits showed that increased force resulted in increased tooth movement; however, the increase in tooth movement rate was proportionally less [31]. While in another study, a 50% increase in force did not register significant differences in tooth movement over 84 days and patients reported greater discomfort at the higher force levels [32]. Furthermore, a 12-week human subject study measuring space closure with light and heavy forces found that initial tooth movement was higher with lighter forces and that the use of heavier forces resulted in undesirable movements, such as loss of anchorage and control [33]. In a large study of 204 dogs, Ren et al. [34] were unable to identify a low force threshold to trigger tooth movement, nor a high force threshold which would indicate an optimum force region. As pointed out by Alikhani et al. [35], increasing the orthodontic force has not been shown to increase the biological response of tooth movement and, as such, high forces should not be used to avoid their harmful side effects.

In the literature, there is an evident lack of consensus on the role of orthodontic force on the rate of tooth movement. A number of reasons could exist for this. Primarily, the orthodontic force system is complicated, as is the biological system on which it is operating. Recent modeling work using the finite element method has shown that the resolution of forces and the stress induced in the PDL is quite complex and difficult to predict [36, 37•]. Furthermore, the magnitude and direction of forces applied to teeth in even the most simple archwire and three-bracket system are difficult to predict [38]. Modeling results of unloading behavior of a wire show that a pseudoelastic Nitinol archwire can express an inverted load-deflection rate where the load increases during deactivation because of reduction in the binding friction between wire and bracket [38, 39]. Clinical studies that have attempted to characterize the effect of force magnitude on tooth movement are affected by uncertainty in the actual force system, case-by-case variation in patient malocclusion, and natural genetic differences in the subject pool. Compounding this is the lack of consensus on the meaning of light versus heavy loads as well as the appropriate time frame for measuring treatment progress. Clearly, it is difficult to interpret the literature when so much noise is present in the data; however, it is consistently found that low-magnitude continuous forces promote efficient tooth movement while lowering the risk of undesirable movements and harmful side effects such as root resorption.

Differential Force Archwires

In the orthodontic literature, it is normal to discuss the relationship between force magnitude and tooth movement; however, it is the distribution of stresses in the PDL that develop as a reaction to the applied loading conditions that generate the biological response and associated tooth movement. A force applied to the central mandibular incisor will generate a much different stress distribution in the PDL compared to the central maxillary incisor due to the significant difference in PDL surface area. A side effect, of a typical archwire replacement progression of increasing cross-sectional area, is that teeth with smaller root structures move first, followed by teeth with larger root structures as wires are replaced with larger cross sections. As the load-deflection rate of modern wires is nearly ideal, the standard archwire progression enables step-by-step correction of malocclusion which is not entirely efficient. Thus, it is desirable to have an archwire with differential properties along its length so that stress can be equalized in the PDL [37•], enabling concurrent tooth movement and reducing overall treatment time.

Several strategies to express differential forces across the arch using a single archwire appliance have been employed. Begg in 1956 [38] proposed the use of differential forces to target the optimum force range for each tooth. He used a thin round stainless steel archwire with a series of loops and bends designed to vary the forces across the arch. In comparison, Burstone [40] proposed a segmented archwire composed of multiple wire cross sections that could be joined together to control the moment to force ratios in different regions of the arch [4]. Both of these methods were time consuming and difficult to accurately control. As the load-deflection rate of orthodontic materials improved with new alloy development, Burstone [41] envisaged a continuous, variable-modulus archwire where the load-deflection rate could be varied; however, the technology to make such a wire did not yet exist and he had to settle for a segmented wire with different materials connected to make a full arch [40].

Fabrication of segmented wires using advanced orthodontic alloys is not trivial. Sevilla et al. [42] presented a segmented pseudoelastic Nitinol and copper NiTi wire that were laser welded to create two segments. They were able to observe two distinct pseudoelastic stress plateaus corresponding to the different mechanical properties of each respective segment. The maximum deformation of the material was reduced by 50% in the welded regions, which is a concern for potential wire breakage during clinical use [14]. Welding materials typically degrade mechanical performance, and this is marked for Nitinol alloys, especially when joining to other materials. Li and Zhu [43] attempted impact butt welding of Nitinol to stainless steel; however, the joint was too weak to allow pseudoelastic stress to be developed in the Nitinol section under loading, and the joint morphology was too rough for biomedical applications. Matsunaga et al. [44] considered both laser welding and electrical resistance welding for joining beta-titanium wires to stainless steel and cobalt-chromium wires. Their results showed comparatively good mechanical properties compared to homogeneous welds in Nitinol or beta-titanium alloys; however, performance against the unwelded base material was not presented. Due to the difficulty in fabricating segmented wires, more attention and success have been paid to development of continuous differential force wires with variable moduli.

As discussed earlier, thermomechanical treatment to affect the TTR of Nitinol alloys can be used to shift the pseudoelastic plateau of the material. When this is done on a local scale within an archwire, differential forces can result. The development of the direct electric resistance heat treatment (DEHRT) process by Miura [45] in 1988 to create wire bends in Nitinol alloys led to the development of continuous differential force wires in 1991 [13, 45]. These archwires were heat treated differently along the arch to achieve a low stiffness in the anterior section, progressing to a higher stiffness in the posterior section, generally with a transition zone between them. Pseudoelastic Nitinol alloy archwires processed in this manner generally exhibit a low-force anterior region where the TTR has been raised through heat treatment, thereby reducing the pseudoelastic stress plateau. The posterior region of the archwire retains the original TTR and pseudoelastic properties while a transition region between the anterior and posterior will have a gradually increasing stiffness [46]. These types of *multiforce* archwires were first commercialized by GAC (Dentsply GAC, Bohemia, NY) and have been in widespread use with minor variations between manufacturers [47]. As pointed out by Mehta [48], the heat treatment process can be expensive and time consuming, and if different heat treatments are required on the same wire, it can be difficult to accurately control the heat and cooling rates, causing high variation in the output. Moreover, there is an apparent limitation on the resolution of the stiffness gradient because of conduction of heat through the material, and only wires with three segments are currently commercially available. On the contrary, varying the material properties as needed in a single archwire would be an advantageous for load resolution, stiffness control, and number of the segments.

A laser process for controlling the pseudoelastic properties of Nitinol alloys with fine resolution has recently been presented [49••]. Selective vaporization of nickel in desired areas reduces its ratio in the chemical composition, thereby altering the material properties. The TTR in these areas is shifted higher in proportion to the amount of nickel removed. This allows for fine control of the pseudoelastic stiffness of Nitinol wires along their length. This process has been applied to copper NiTi archwires for use in orthodontics. These laser-processed wires are effectively programmed along their length so they deliver the desired differential load at any location. Using the ideal force proportions developed by Viecilli and Burstone [37•], an archwire was developed to express target forces for each tooth in both the mandibular and maxillary arches. The benefits of the selective laser vaporization approach compared to the selective heat treatment approach outlined above are that a higher resolution of stress gradients and greater range of force levels are possible. While each approach maintains the ideal load-deflection rate and large activation ranges of pseudoelastic nitinol, selective laser vaporization varies forces on a tooth-by-tooth basis, while heat treatment is limited to a gradually decreasing stiffness from the posterior to anterior segments. The impact of this difference becomes apparent when considering the differences between the PDL and interbracket distance (IBD) between adjacent teeth. The mandibular lateral incisor has a much smaller PDL surface area and less IBD compared to the mandibular canine, so each tooth requires a vastly different archwire stiffness to deliver an ideal load. Using the laser process to program the desired stiffness enables a long-lasting archwire to deliver a light, continuous load to each tooth in the arch. Concurrent tooth movements can be attained with a single archwire, accelerating treatment kinetics while avoiding potential side effects, such as the delayed tooth movement that is associated with progressive archwire therapy [49••, 50].

Applications beyond Orthodontics

Nitinol has long been considered as a suitable material for medical devices and surgical applications due to its biocompatibility and desirable shape memory and pseudoelastic properties [8, 9]. As early as 1976, Nitinol was considered for Harrington rod treatment of scoliosis, when Schmerling et al. [51] placed martensitic rods in cadavers, which were then heated using electrical resistance to straighten the rods. Research in the use of shape memory alloys has continued over the years. Pseudoelastic wires for gradual scoliosis correction were recently investigated by surgically implanting orthodontic wires in rats [52]. Gradual post operative correction is advantageous for avoiding complications such as neurological injury which may occur with instantaneous correction. The use of tailored rods for customized load applications to different areas of the spine is an attractive research topic with substantial clinical potential.

One of the first medical devices that utilized Nitinol was the bone suture anchor. This device has been popular for Bankart repair procedures in the shoulder [53, 54], as well as for the fixation of temporomandibular joint devices [55–57]. Research continues to produce good results utilizing Nitinol bone suture anchors for new procedures such as an alternative to K-wire fixation [58]. Bone fixation devices, such as orthopedic bone staples, have also been a good application of Nitinol [59, 60]. Nitinol staples have produced consistently favorable bending stiffness for immobilization and fusion across fractures or in arthrodesis procedures [61, 62]. In these cases, the dynamic flexibility of the Nitinol material can offer compressive stress and permit earlier weight bearing after a procedure. Further research, especially in mathematical modeling, such as the work done by Saleeb et al. [63], would be useful in determining the suitability of differential force technologies in these applications.

Conclusions

The application of light, continuous forces over the full activation range continues to be an important function of the orthodontic archwire. Nickel-titanium-based alloys, such as pseudoelastic Nitinol and copper NiTi remain the preeminent materials for delivering these continuous forces over extended ranges due to their pseudoelastic properties and ideal load-deflection rates. At lower activation deflections, a relatively new beta-titanium alloy known as Gum Metal delivers exceptional low load-deflection rates and is suitable for orthodontic applications, especially in the finishing phase.

The fabrication of differential force wires with variable stiffness along their length is an increasingly attractive process. Conventional methods for fabricating composite archwires with variable properties are limited by the poor mechanical properties of the welded joint. Heat treatment processes can be used to create continuous archwires with modestly variable force to deflection characteristics for up to three different regions. The lack of resolution and limited force variability of selectively heat-treated wires are a weakness that limits their clinical application. A new process using selective vaporization of nickel can resolve this weakness by controlling the wire stiffness with a very high spatial resolution. This allows each tooth in the arch to be targeted with loads proportional to the PDL area, even when the IBD is limited. Advanced Nitinol technology is increasingly attractive for medical and surgical applications outside the field of orthodontics.

Compliance with Ethical Standards

Conflict of Interest Ibraheem Khan has a patent (US20170224444A1) pending, and Smarter Alloys Inc. (with which all authors are affiliated) is involved in the design and manufacture of appliances for orthodontic treatment.

Human and Animal Rights and Informed Consent This article does not contain any studies with human or animal subjects performed by any of the authors.

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