# Advanced surface engineering technology for endodontic instruments and related applications

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**Fig. 1**_Relation between Rockwell hardness and microhardness from steels to hard ceramics and superhard materials.


## _Summary

Application of physical and chemical vapor deposited surfaces (PVD and CVD) affects the cutting efficiency, corrosion and abrasion resistance, and biocompatibility of a wide variety of dental instruments, including NiTi and stainless endodontic files and reamers, scalers and curettes, ultrasonic tips, scissors, scalpels, implant drills, and various types of stainless and carbide burs. The unique patented Large Area Filtered Plasma Deposition (LAFPD) technology offers surfaces of virtually unlimited compositions and architectures deposited atom-by-atom on complex shaped substrates made of different materials: from certain plastics, to stainless steel, to carbides and ceramics. In addition, this process is capable of pre-deposition ion plasma treatment of substrates by modifying the surface layer with different alloying elements.

The LAFPD process is capable of forming surfaces from materials which cannot be created by conventional metallurgical processes. Among such materials are super hard diamond-like (DLC) and related surfaces, nanocomposite and multilayer metal-ceramic films, continuous polycrystalline diamond surfaces and many more. The combination of pre-deposition and post-deposition treatment with optimized surfacing allows substantial improved service life and performance of dental instruments. The role of surface engineering against wear, corrosion, friction and fatigue behavior of endodontic and other dental instruments will be discussed. Special attention will be paid to endodontic NiTi

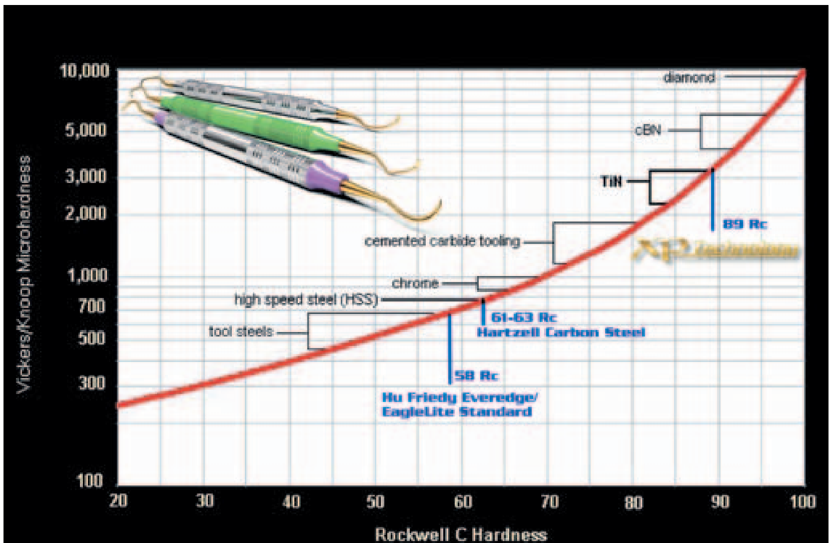




Fig. 1

and stainless files and reamers, endodontic ultrasonic tips, endodontic drills, and endodontic burs, as well as other dental related products and applications.

## _Introduction

Interaction between dental instruments and different kinds of dental tissues affects both tool performance and the response of the dental tissue. The highly corrosive and abrasive environment associated with treatment of hard dental tissues such as enamel and dentin, as well as soft tissues, can affect the cutting performance of an instrument by changing the geometry of the cutting edges. It also affects the profile and surface chemistry of the metal surface, resulting in changes of friction and galling properties during operation. Conversely, the changes in the instrument performance have a biomedical consequence on the dental tissues. Using an instrument in a sub-optimal condition can result in short- and long-term post-operative consequences ranging from potential overheating of tissue, contamination of the tissue by transferring the metal elements from the instruments to the dental tissue, increased bleeding, allergic reactions and inflammation.

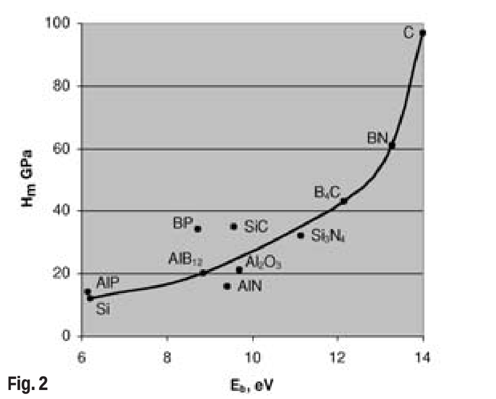



Fig. 2

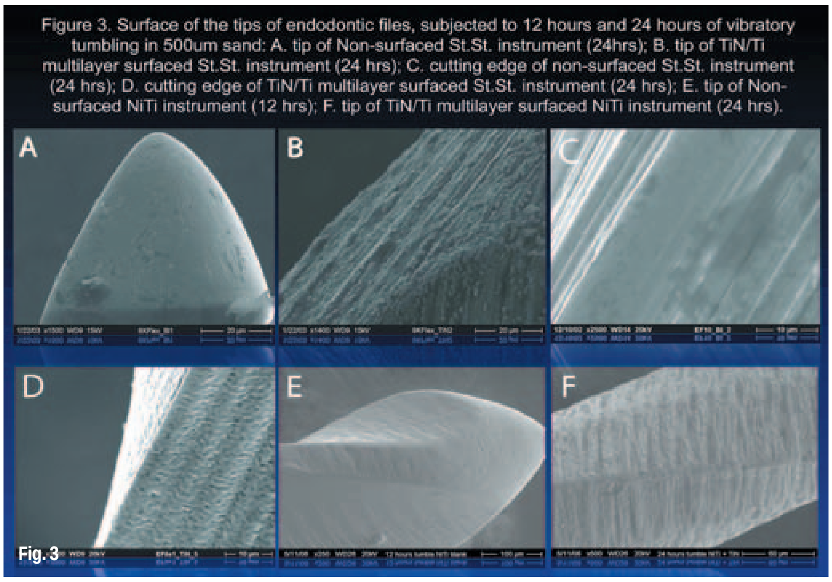
In most dental applications, instruments must provide the following properties: cutting efficiency, long-term cutting edge retention, ductility, abrasion and corrosion resistance, long fatigue life (this is especially important for rotary instruments), biocompatibility, and anti-galling properties; with no transfer of metal from the instrument into the tissue. Bulk materials (metal, ceramics, plastics, composites) cannot always provide all the desired properties of the instruments required for their use in different dental and medical applications. Therefore, surface engineering can add some important advantages to an instrument. In some cases, surface engineering is used to improve the existing properties of the instrument; in other more important cases the bulk metal serves as a carrier for the carefully designed surface structure, which provides important functional properties for selected applications.

Abrasion resistance and cutting edge stability are important characteristics of a dental instrument and are closely related to the hardness of the material of the instrument. Rockwell hardness is widely used for the measurement of hardness of metals. It is determined by the residual plastic deformation left on the surface of a material after indentation by the indenters of various geometries at certain loads. It employs loads ranging from 100 grams to several kilograms. In the case of hard and super hard ceramic materials such as carbides, nitrides, and borides, the Microhardness methodology replaces the Rockwell technique. The Rockwell technique cannot be used for hard and brittle ceramic materials, which will be destroyed by large indentation forces. Therefore, Microhardness has to be used for assessment of the hardness and other mechanical properties of hard ceramic materials. Microhardness measures plastic deformation of materials under indentation by a diamond pyramidal indenter of different geometries (Vickers, Knoop, Berkovich). As small as 1 to 10 grams are used in the test, which results in a characteristic indentation site of a few microns (1 µm= 1/1,000 mm or 1/25,000 inch).

The relation between Rockwell hardness and Microhardness of different materials is shown in Figure 1. It can be seen that hardness is a strongly non-linear parameter. There is a qualitative difference between hardness of hard steels or cemented carbides and super hard materials such as boron carbide, cubic boron nitride and diamond. This is

**Fig. 2**_Hardness of materials vs. bonding energy of atoms in the lattice.

**Fig. 3**_Surface of the tips of endodontic files, subjected to 12 hours and 24 hours of vibratory tumbling in 500 um sand: **A.** tip of non-surfaced stainless steel file (24 hrs): **B.** tip of TiN/Ti multilayer surfaced stainless file (24 hrs); C. cutting edge of a Non-surfaced stainless steel file (24 hrs); **D.** cutting edge of a TiN/Ti multilayer surfaced stainless steel file (24 hrs); **E.** tip of non-surfaced NiTi instrument (12 hrs); **F.** tip of a TiN/Ti multilayer surfaced NiTi file (24 hrs).



a result of the different energies of atoms in the lattice of different materials as shown in Figure 2.[1] The drawback of hard and super hard ceramic materials is their brittleness. It is well known that even super hard crystalline materials such as boron carbide or diamond have the highest abrasion wear resistance, but they are extremely brittle and can be easily fractured by impact or direct pressure. Contemporary surface engineering technology is capable of combining the elastic properties of metals and the high hardness of the super hard ceramics to achieve super hard, but also extremely tough surfaces. This is achieved by applying hard ceramic or cermet (metal-ceramic) surfaces on the surface of metal components. In the case of cutting tools, the hard and super hard ceramic surfaces protect the cutting edges against abrasive wear.

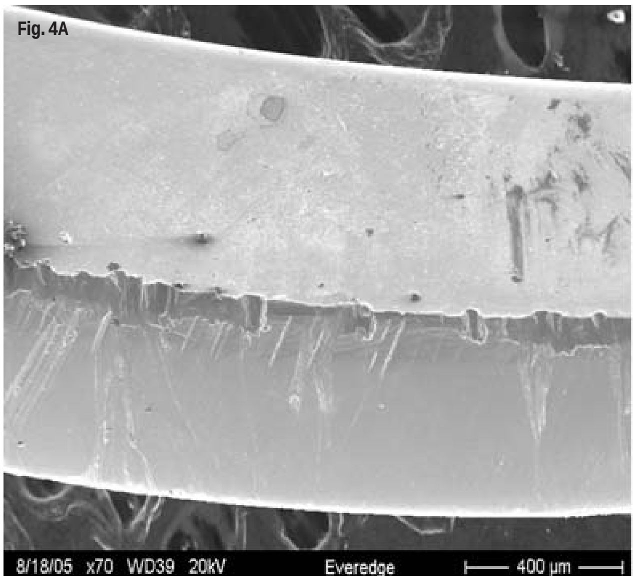
## _Surface engineering in dental applications

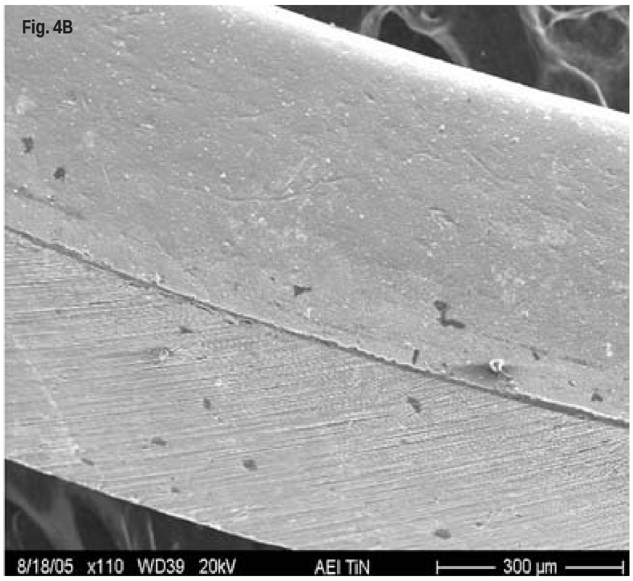
_Endodontic The increase of abrasion resistance of K-files and nickel-titanium root canal files by application of a titanium-nitrite (TiN) surface by advanced large area filtered cathodic arc deposition (LAFAD) technology was investigated at American Eagle Instruments, Inc. The instruments, of different grades, were subjected to vibratory tumbling in 0.5 mm size quartz sand for 24 hours. The results shown in Figure 3 have demonstrated that non-surfaced instruments, both stainless steel and nickel-titanium (NiTi), are completely polished; all cutting striations have disappeared and the cutting edge is flat after tumbling. In contrast, the TiN surfaced instruments demonstrate either no change, or no noticeable change in their surface profile or in the cutting edge sharpness. These results support the general statement that soft material cannot affect hard materials by abrasive interactions. Going up along the hardness chart curve presented in Figure 1, it is established that harder materials cannot be affected (abrasion wear) by abrasion media composed of a softer material. This explains the results presented in Figure 3: quartz sand cannot scratch the TiN surface because this surface (2,500–3,000 Vickers) is much harder than quartz (1,200 Vickers). Both surfaced stainless steel files (Figs. 3b,d) and NiTi files (Fig. 3f) with a Ti/TiN multilayer cermet surface deposited by the LAFAD process demonstrate little detectable changes in the surface pattern after 24 hours of vibratory tumbling with an abrasive media of quartz sand. Conversely, non-surfaced files show an entirely polished pattern of wear with complete dulling of the cutting edges.

_Periodontal Using hard and super hard surfaces can dramatically change the cutting performance of many types of dental instruments such as endodontic files and reamers, curettes and scalers, scalpels, implant drills and burs by retention of the cutting edges. It has been successfully demonstrated in dental practice with curettes and scalers with hard titanium nitride (TiN) surfaces recently taken to the marketplace by American Eagle Instruments, Inc. When an instrument with a TiN surface scrapes the cutting edge of another instrument made of hard steel, the result is always the same: there is no material transfer from the hard ceramic to the steel. The opposite is true in that the TiN surfaced instrument easily takes metal from the cutting edge of the steel curette shown in Figure 4.

The cutting efficiency and cutting edge retention of periodontal dental instruments were inves-

**Fig. 4_ A.** Hu-Friedy Everedge Curette cutting edge after abrasion against AEI LAFAD TiN surfaced Curette's cutting edge; **B.** American Eagle Curette cutting edge after scraping against Hu-Friedy Everedge Curette's cutting edge.



tigated both in laboratory testing and in dental practice. In laboratory tests the TiN surfaced and non-surfaced curettes and scalers were subjected to stroke tests against bovine bone at controlled loads.[2] The metallographic cross-sections of the cutting edges of the curettes and scalers after a certain number of strokes are presented in Figure 5. It shows the results of using photomicrographs of the cutting edge, a cross-section in order to assess wear of surfaced and non-surfaced instruments. Comparison of the photomicrograph of the cross sections of the instruments at a representative point on the blade demonstrates the wear of non-surfaced instrument after 1,500 strokes (Fig. 5A) versus 15,000 strokes of the surfaced instrument (Fig. 5B).

Further investigation of the cutting edge retention was done at the University of Toronto, Institute for Biomaterials of the School of Dentistry (Professor P. Watson). SEM micrographs of cutting edges were made at magnifications of 100 and 1,000 after 0, 500, 2,000 and 5,000 cycles of sliding contact against bovine enamel.[2] The 100x magnification micrographs were used to construct a composite graphic along the length of the cutting edge of the blade to assess whether wear was uniform over the length of the blade. Wear was determined by measuring the width of the cutting edge of the instruments at four stages: as received from AEI, after 500, 2,000 and 5,000 cycles. Instruments were coded so that any differences in surface variables between surfaced instruments were unknown during the evaluation process. The bar chart (Fig. 6) shows widths of the cutting edge for each instrument in microns measured from SEM photomicrographs made at a magnification of 1,000 power. The instrument codes are shown on the horizontal axis. The first seven instruments were surfaced. The last four instruments were not surfaced. All the surfaced instruments showed an increase in the width of the cutting edge from approximately 1 micron before testing to an average of 3.4 microns after 5,000 cycles. The control (non-surfaced) instruments showed an increase in the width of the working edge from approximately 1 micron, before testing to an average of 35 microns after 5,000 cycles. The non-surfaced blades showed 10.3 times more wear than the surfaced ones.

The example of cutting edge retention of periodontal curettes in dental practice is shown in Figure 7. In this case, the cutting edge width did not exceed 40 µm after 11 months of intense usage. These instruments, having hard ceramic surfaces do not require re-sharpening, while conventional instruments made of hard steel require re-sharpening practically after the treatment of each patient.

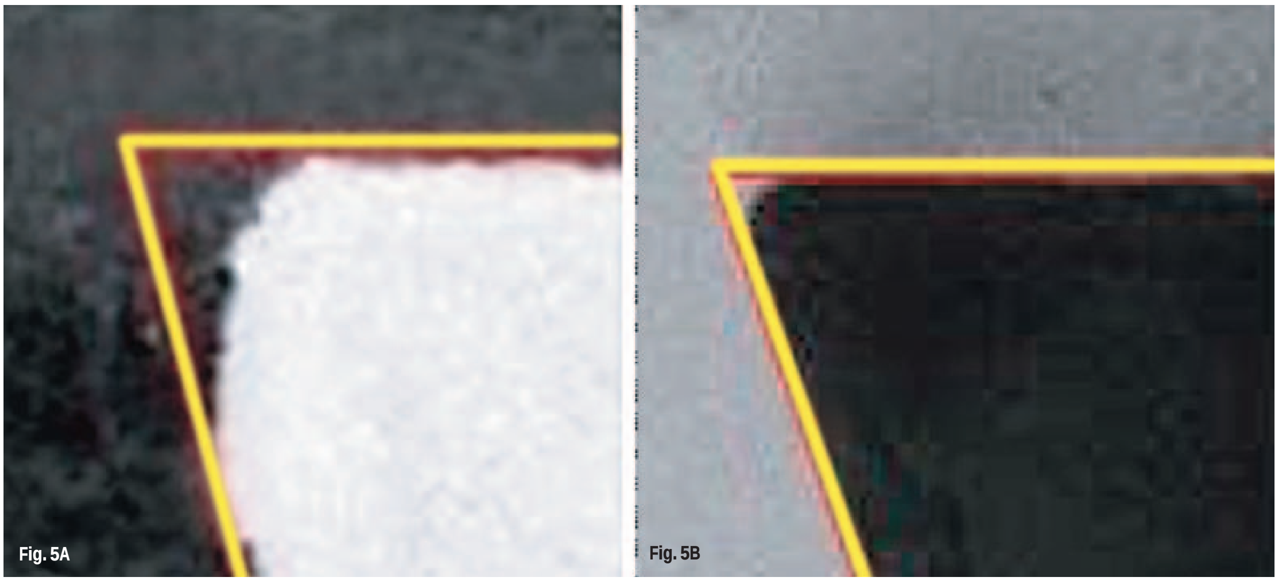



**Fig. 5**_ Metallographic cross-section after stroke testing. **A.** 1,500 strokes with non-surfaced scalers. **B.** 15,000 strokes with Ti/TiN multilayer surfaced scaler.

_Ultrasonic Recently developed surface engineered ultrasonic scalers at American Eagle Instruments have demonstrated more than a 20-fold increase in service life. The wear land developed on the surface of a surfaced ultrasonic scaler after 120 hours of interaction with bovine at fixed load of 83 gram shows only polishing abrasion wear (Fig. 8). In comparison, the non-surfaced ultrasonic scaler failed after 6 hours by developing a deep groove at the operating contact area. Similar tests using Piezo type ultrasonic scaler tips under the same conditions showed 2 mm wear of the tip at 35–38 hours of operation. The surface engineered

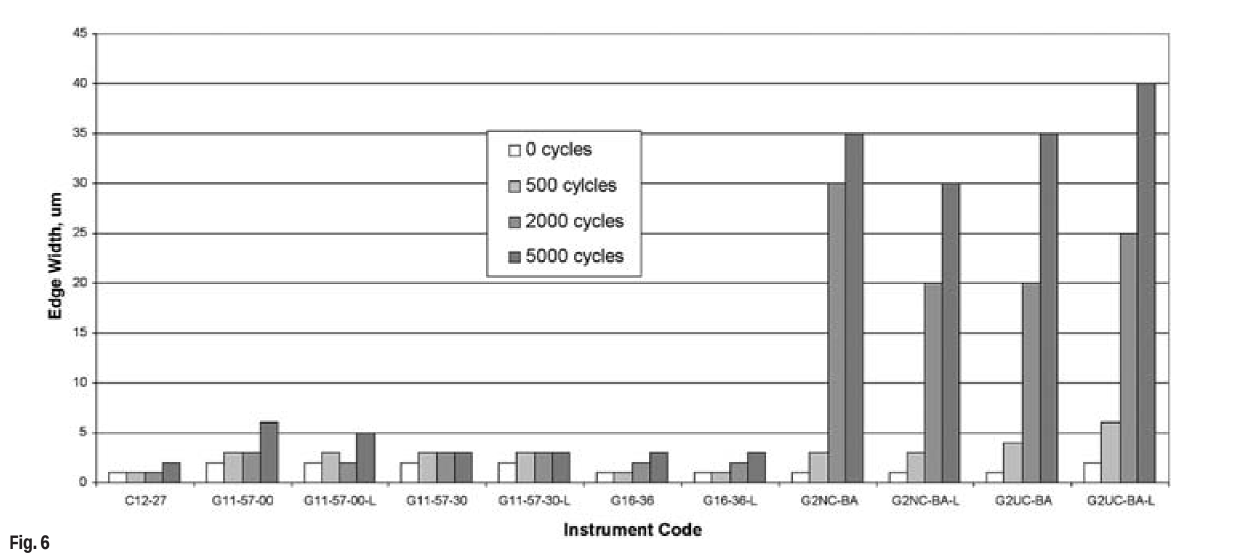



Fig. 6

**Fig. 6**_Chart representing wear of different instruments obtained by stroke testing at University of Toronto Faculty of Dentistry.

surfaces have not shown any measurable wear after over 145 hours of test operation.

_Rotary Tools When this type of TiN surface is applied to rotary instruments it results in a dramatic increase of cutting efficiency. In one instance the TiN surface was applied to implant drills. Testing was preformed that involved measuring the depth of a drill hole in a bone after a certain numbers of revolutions (Dr N. Bekesch). As expected, the cutting ability of TiN surfaced drills is much higher than non-surfaced ones (Fig. 9). As the cutting edge wears, its ability to bite into the substrate diminishes and the cutting rate (distance penetrated per revolution) gradually falls to such a low level that the drill stops cutting. As the cutting efficiency decreases, the practitioner will compensate by increasing the pressure on the drill. This increases heat production, resulting in further damage of both the instrument and the tissue (counterpart). Another consequence from the dulling of the cutting edges on the drill flute is that it fails to self-align, resulting in changing the geometry of the implant hole, making it much more difficult to produce a round hole (Professor P. Watson). It is also not known how much of the substrate (drill) material remains in the tissue area or is removed by irrigation or is ingested by the patient.

## _Surface sngineering of NiTi in endodontics

Lack of cutting efficiency is a substantial drawback in nickel-titanium (NiTi) root canal files. Nonetheless, its extraordinary shape memory properties allow the instrument to stay in tight eccentric contact with the surrounding walls of the root canal. The low hardness of NiTi (20–30 Rc), which is much softer than stainless steel (53–63 Rc), results in a rapid degradation of the cutting edges.[11] To sustain the necessary forward movement of the tool requires an increase in pressure on the file, which can often result in instrument separation. NiTi is also known for its high friction propensity. In rotary operation, this may result in increased heat generation as well as excessive damage to the root canal tissue if the doctor is not careful. Depositing a hard ceramic or cermet surface on NiTi instruments can effectively reduce friction, improve wear resistance of the


Fig. 7A

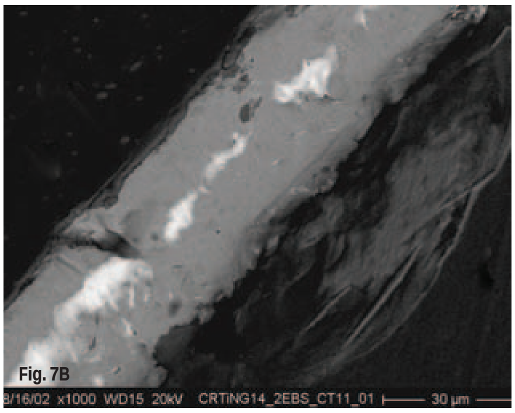



Fig. 7B

**Fig. 7**_Wear of TiN surfaced scalers in clinical trials. **A.** intact edge after 11 patients, **B.** wear land after 11 months in clinical usage.

cutting edges and substantially increase cutting efficiency. The LAFAD surfacing of NiTi and stainless steel files with low friction, anti-galling hard cermet or diamond-like coatings is performed at low temperatures, potentially without detrimental effect on the bulk metal properties of the instrument. This allows improvements in the cutting efficiency without loosing the shape memory or spring characteristics of the NiTi and stainless metal alloy.

Testing was performed at American Eagle Instruments on surfaced (multilayer Ti/TiN cermet surface) and non-surfaced NiTi endodontic files. Comparison of the cutting edges was made using SEM micrographs of the NiTi endodontic files after drilling a number of holes in bovine bone at 400 RPM (Fig.10). It can be seen that the surfaced files show no changes both on the surface of the flute and along the cutting edge after drilling 25 holes, while the cutting edges of non-surfaced files are completely rounded after drilling 10 holes.

detrimental to fatigue performance, reducing fatigue life up to 80% in comparison with non-surfaced samples.[5] In electroplating, the cracks were initiated at the coating-substrate interface, where residual compression stresses are maximal. This is followed by fast propagation toward the coating surface. When the coating is fractured, the cracks starting the fast propagation continue into the bulk metal and result in reduction of the fatigue life performance of the part. The opposite is true when TiN surfaces were applied. Having high hardness, these surfaces slow the crack propagation and result in a substantial increase of fatigue life.[6] This example demonstrates how important it is to minimize or eliminate surface defects.

The difference between conventional cathodic arc surfaces deposited by direct cathodic arc sources and LAFAD (see below) filtered arc deposition technology is shown in Figure 11. It can be seen that conventional surfaces have a large number of inclusions and voids resulting from

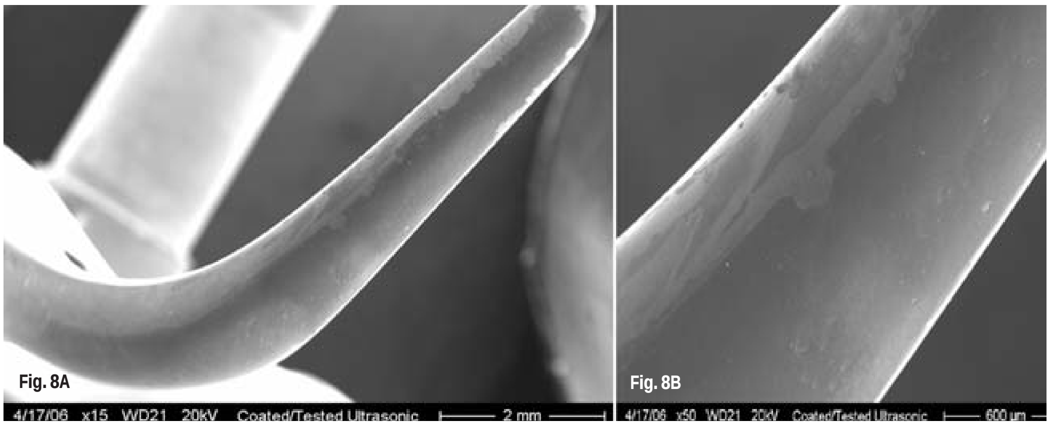



**Fig. 8**_Polishing abrasion wear on a TiN-surfaced ultrasonic scaler developed during test run against bovine at a fixed load of 83 grams for 120 hours: **A.** X15 power **B.** X50 power.

One of the most important properties which contribute to the durability of the endodontic root canal instruments is torsion fatigue. Two major fatigue modes found in NiTi endodontic rotary instruments are ductile mode and brittle mode.[13] In the ductile fatigue mode, micro cracks develop inside the bulk metal of the instrument, starting where bubbles and voids are created by the partial oxidation of the metal during manufacturing of the instruments. Obviously, surface engineering cannot help with this type of fatigue. Brittle fatigue is always initiated at the surface of the instrument, starting from micro cracks, striations and other surface defects. Using defectless hard surfaces it is possible to significantly suppress, if not completely eliminate, this type of fatigue. Hard ceramic surfaces, preferably having multilayer architecture, can help to battle brittle fatigue in cyclic bending. In fact, application of relatively thick electroplating hard chrome or nickel-phosphorus coatings was found to be

macro particles created by evaporation of the cathode target in direct optical contact with the substrates to be coated. The LAFAD technology uses electromagnetic filtering of macro particles by imposing the curvilinear magnetic field on the metal–gaseous vapor plasma flow. This allows complete elimination of any atom clusters or macro particles generated by the primary cathodic arc plasma sources. This technology is capable of producing 100% ionized and atomized metal vapor plasma that comes out of the LAFAD source to be deposited on the instruments. As a result, an atomically smooth, hard TiN cermet surface having a thickness up to 7 mm can be deposited.[2] The LAFAD process is described in more detail elsewhere.[2–4]

There have been a number of attempts to use surface engineering technologies for improvement of endodontic root canal files. Ion implantation was used to improve corrosion resistance

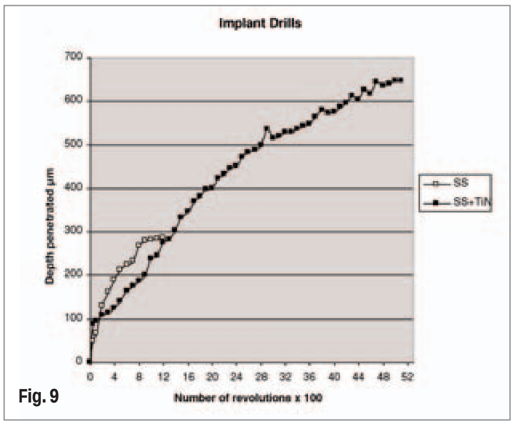



Fig. 9

**Fig. 9**_Implant drills testing result: SS-non-surfaced drill; TiN- drills with Ti/TiN multilayer cermet surface deposited by LAFAD surface engineering process.

and friction behavior of NiTi files and wires.[7,14,18] It was found that forming TiN or TiOx/TiN coatings on the surface of NiTi results in an improvement of biocompatibility, corrosion resistance and reduction of friction. It was mentioned that one of the reasons of using ion implantation instead of depositing a TiN surface was a problem with the interface between the surface and bulk metal.[7] Poor adhesion of conventional TiN coatings deposited by direct cathodic arc evaporation, as well as a high density of defects in these coatings (inclusions, voids, porosity) created by the large amount of macroparticles or droplets coming from the evaporating surface of the cathode target are possible reasons for the low cutting efficiency improvement of TiN surfaced NiTi files reported in.[8] Another reason to use ion implantation instead of surface deposition for improvement of corrosion and wear-corrosion behavior of NiTi is that ion implantation is carried out at near substrate temperatures and, therefore, will not alter the microstructure and bulk properties of the alloy (negatively affect its shape memory).[15] Thermal management of the substrate materials, especially of NiTi instruments and files of small sizes and diameters, is still a challenge to the industry. Failure to manage substrate temperatures of NiTi results in the loss of its shape memory and negatively impacts its resistance to cyclic and torsional fatigue failure.

## _Advancement of surface engineering

Further advancement in LAFAD technology and hybrid Large Area Filtered Plasma Deposition (LAFPD) technology recently developed at Arcomac Surface Engineering, Inc., combine virtually all low-pressure plasma deposition processes such as thermal evaporators, electron beam evaporators and magnetron sputtering, together with LAFAD dual arc plasma sources in one universal chamber layout.[12] Today's LAFPD surface engineering technology is capable of overcoming the drawbacks of conventional surface deposition processes.[2–4] LAFAD is able to deposit a wide variety of surface compositions and architectures by mixing elements in highly ionized plasma and depositing them on complex shape substrates atom-by-atom according to whatever coating design is desirable. Virtually any element of the periodic table can be generated by different vapor plasma sources, integrated in one universal chamber layout of Arcomac's LAFPD surface engineering system. Both LAFAD and LAFPD surfaces are known for their extremely high adhesion, which in some cases even exceeds the coating materials cohesion. The SEM image of a sheared scaler shank with a TiN multilayer surface (3 µm thickness) is shown in Figure 12. It shows the cracks developed through the surface as a result of large plastic deformation of the substrate. However, it shows no delaminations up to the edge of the sheared area, demonstrating the super adhesion properties of LAFAD cermet surfaces. It has to be noted that conventional TiN surfaces in the same conditions display catastrophic delamination (lack of cohesive strength) around the edge of the sheared area.

In surface engineering of dental instruments, adjustment of the hard surface properties of the surface engineering to the relatively soft substrate metals can be a challenge. It can create a so-called "egg-shell effect" when a hard ceramic surface lays on a soft substrate base that can be easily deformed by loads perpendicular to the surface. A duplex technology has been developed to address this problem. In duplex technology, a relatively thick interlayer is created between the hard surface and soft metal base. This interlayer has a medium hardness between the hard coating and the soft metallic substrate. This layer can be made by a number of different techniques: ion nitriding, carburizing, ion implantation or hard electroplat-

**Fig. 10**_Comparison of the NiTi endofiles after drilling test (400 rpm).
**A:** Cutting edge of uncoated instrument after drilling 10 holes,
**B:** Cutting edge of TiN coated instrument after drilling 10 holes,
**C:** TiN coated instrument after drilling 25 holes,
**D:** Cutting edge of TiN coated instrument after drilling 25 holes.

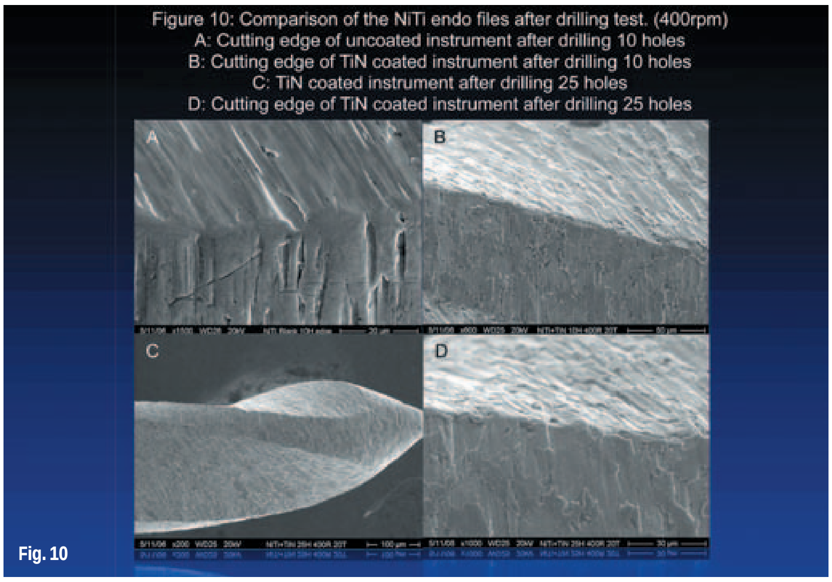



Fig. 10

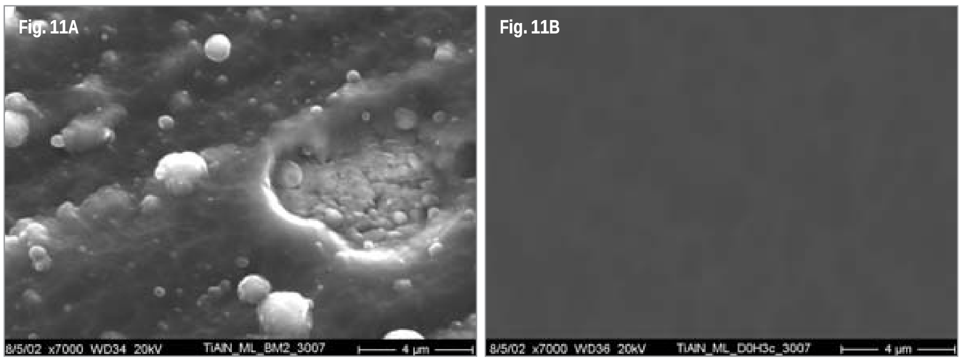



**Fig. 11**_Competitive surfaces with Equal Magnification and Thickness: LAFAD vs. Conventional Direct Cathodic Arc Deposition (DCAD).
**A.** "Kennametal PVD Applied Thin Film"
**B.** "Arcomac 'Filtered Arc' Applied Thin Film"

ing coatings. The intermediate layer will smooth the sharp gradients between the hard ceramic thin film surface and the soft metallic substrate. LAFAD technology is capable of making ion nitriding or ion implantation treatment almost as precise as the thin film super hard surfaces it can create. The thickness of an ion nitrided layer prepared by LAFAD plasma immersion ion nitriding process in strongly ionized nitrogen plasma can range from 3–5 µm to 100 µm with +/–1 µm in accuracy. In the LAFPD surface engineering system all the technological steps including ion cleaning, ion nitriding or ion implantation and a number of multilayer surface deposition steps can be performed in one vacuum cycle without the need to transfer parts from one chamber to another. Complete surface engineering technologies also include pre-deposition and post-deposition treatment. Pre-deposition treatment includes different finishing procedures such as chemical-mechanical polishing, electro-polishing and vibratory tumbling. The post-deposition treatment may include different heat treatment and cryogenic treatment techniques.

The capability of LAFAD technology to deposit atomically smooth, optical quality surfaces onto complex shape parts in large batch load processes is equally important for corrosion resistance and biocompatibility of the instruments. It is well known that pitting corrosion damage often starts at surface defects in the substrate material. When the initial metal surface is mirror smooth, surface defects become the secondary points where corrosion starts. LAFAD surfaces not only eliminate this avenue of corrosion but also substantially reduce the primary source of corrosion by covering any surface defects, and filling surface voids and imperfections. The LAFAD plasma source generates 100% ionized metal vapor plasma flow having substantial kinetic energy ranging from 40 eV to 200 eV. When the flow of energetic metal ions impacts the substrate surface it mixes with the substrate atoms forming a transitional layer, which can secure excellent adhesion even at room temperature of the substrate material during the plasma processing. The LAFAD process is also capable of precise thermal management of substrate temperatures by controlling the magnetic shuttering of the plasma flow. In this case the depositing metal vapor plasma stream is periodically deflected from the substrate material. During a pause in the surface deposition process the thermal flux is evenly distributing into the bulk substrate's body, which prevents overheating the edges and allows us to precisely control the integral temperature of the substrates. Using this approach the TiN surfaces were deposited on tiny medical instruments made of stainless steel and titanium alloys that resulted in an order of magnitude improvement in their corrosion resistance.[16]

To meet the challenges of modern dental instrument technology, surface engineering has to be capable of providing not only extremely smooth and defectless surfaces, but also multi-elemental multiphase material with sophisticated architectures. One example of such a surface is a multilayer architecture with a metallic layer (Ti) followed by a ceramic layer (TiN) as is shown in Figure 13. This surface design allows combining extremely hard ceramic layers with plastic-like soft metal layers to create a flexible yet hard film on the surface of metal instruments, which provides the required ductility and elastic response needed in dental operations. Multilayer surfaces have a unique capability to divert microcracks from penetrating into the bulk substrate metal as illustrated in Figure 14. As illustrated, the micro cracks initiated on the instrument surface are arrested in the multilayer surface structure, which prevents it from further propagating into the bulk metal. When the surface is engineered to have substantial compressive stress, the cracks are often collapsed even before they have a chance to start to propagate. An advanced development in this

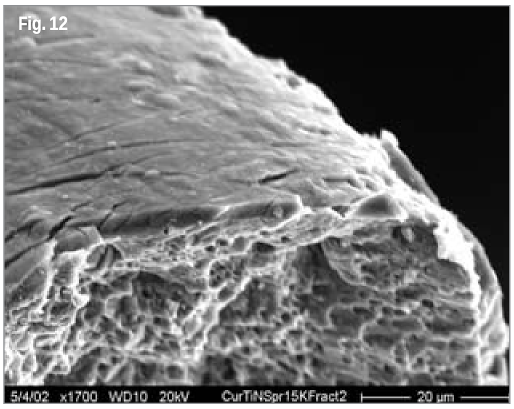



**Fig. 12**_SEM image of sheared scaler shank. Scaler has 3 µm TiN/Ti multilayer surface.

direction is nanolaminated coating architectures.[10] An example of a nanolaminated CrAlN coating deposited by LAFAD process is shown in Figure 15. In this type of surface each sublayer has a thickness of 1–3 nanometers or 10–30 Angstrom.[9] For comparison, the diameter of a water molecule is about 3 Angstrom. Each layer of this surface is composed of nanocrystals having dimensions as small as 10 Angstrom. When such ultra-fine crystals are embedded into an amorphous matrix made of diamond-like carbon, a very unique combination of super hardness with high surface toughness can be achieved.[10] The nanostructured and diamond-like surfaces have demonstrated dramatic improvements in abrasion and abrasion-corrosion behavior of biomedical stainless steels and their corresponding products.[17]

## _Biocompatibility of nickel-titanium endodontic files

The goal of a medical device is that it maximizes its usefulness in its defined application while minimizing the detrimental effects both locally and systemically. Biocompatibility has been equated with inertness; however, it is clear our thinking must expand this definition. Williams[19] defines biocompatibility as “the ability to perform with an appropriate host response in a specific situation.” In other words, it is important not only to examine the interaction between the instrument/implant to the tissue, but also the effect of the tissue on the instrument/implants performance. With this in mind, what is required of a nickel-titanium endodontic file?

The endodontist desires an efficient, effective tool that reams the canal, clears all soft tissue, widens the canal in the dentin and leaves a smooth, tapered track. The file must be flexible enough to follow the tortuous canal, but tough enough to withstand the torsional fatigue and maintain its sharpness against the very abrasive dentin while retaining its inertness to the tissues.

The advantages of the current nickel-titanium endodontic files are well established, but so are the deficiencies. These disadvantages include the difficulty in manufacturing to produce a smooth, sharp edge.[20,21] Because the cutting edges of nickel-titanium files have a micro hardness of 303 to 362 Vickers units, they are extremely susceptible to dulling and wear.[22] These files are also susceptible to a decrease in cutting efficiency with repeated sterilization.[23] Once the file falls into a sub-optimal condition, the instrument’s performance will have biomedical consequences on the procedure. This includes potential overheating of the tissue with possible tissue necrosis, failure to create a smooth canal, elongated procedure times and contamination of the tissue with metal elements from the instrument. Using an endodonic file that is dull or compromised decreases the efficiency of the endodontist, increases the risk of instruments failure to perform adequately and possibly fracturing of the file within the tooth.

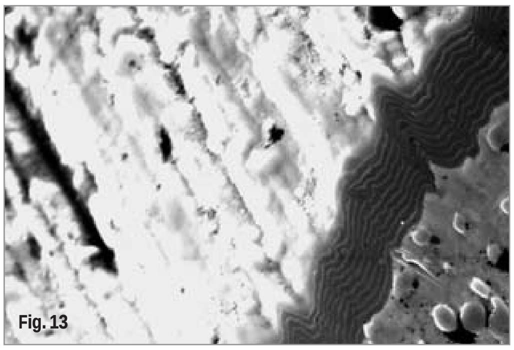


**Fig. 13**_SEM image of scaler with TiZrN/TiZr multilayer surface 2.5 µm thick after 9,000 strokes. The multilayer structure having TiZrN ceramic layers 0.3–0.4 µm thick followed by TiZr metallic interlayers 0.03–0.05 µm thick is revealed across the wear land.

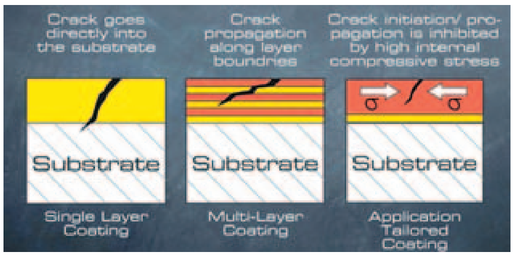



**Fig. 14**_Schematic illustration of crack propagation in different surface architectures (Courtesy of Dr. Zimmerman).

Nickel-titanium is considered safe with acceptable compatibility to tissue mainly due to the formation of titanium oxides at the surface. However, continued concerns exist with the dissolution of nickel (Ni) ions with the possibility of allergic[24,25], toxic[26] and carcinogenic[27,28] effects. Much of the toxicology research has been done on static implants and not on dynamic instruments like the endodontic file. Stress aggravates ion release[29], along with the fact that mechanical damage to the instrument causes continued virgin surface exposure and may have increased local release of Ni. Little is known of the physical effects of metal shavings left behind from drills and files as they wear. Orthopedic literature is clear on the roll of implant loosening from metal and polyethylene particles that occur during wear. The effects of Ni released systemically are not clearly understood. One concern is late sensitivity to Ni after initial exposure that could induce sensitivity to future dental, orthopedic or cardiac implants.

Surface engineering with titanium nitride and other hard and super hard cermet surfaces addresses and improves each of these concerns. It minimizes or eliminates the exposure to the nickel and its toxicity, while improving the mechanical properties of the file itself._

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**Fig. 15**_CrAlN nanolaminated surface deposited by LAFAD surface engineering technology: **A.** nanolaminated surface architecture, representing near 1000 layers per one micron of surface thickness; **B.** nanocrystalline structure of surface sublayers.

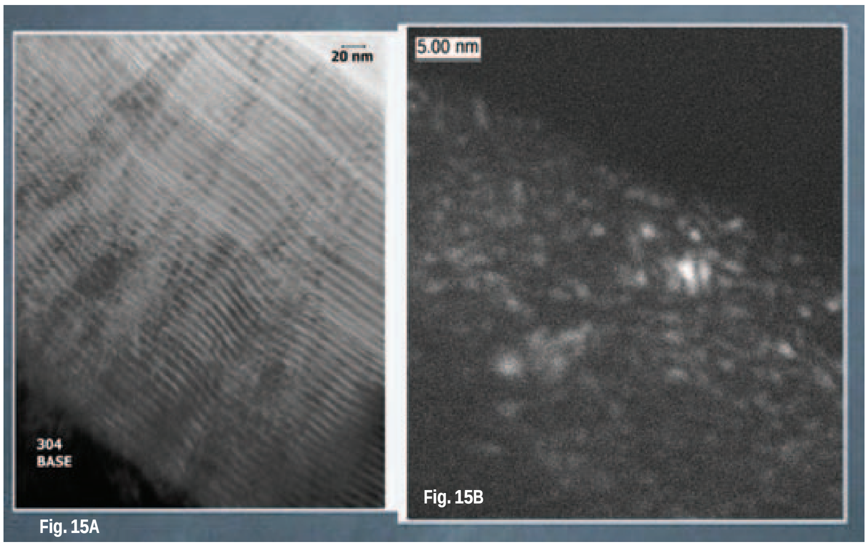


Fig. 15A

Fig. 15B

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