Predicting *In Vivo* Failure of Pseudoelastic NiTi Devices under Low Cycle, High Amplitude Fatigue

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Abstract: Due to the large reversible strains achievable through the stress-induced austenite–martensite phase transformation in NiTi alloys, NiTi has replaced stainless steel in the majority of large-strain biomedical applications such as root canal enlargement. However, the pseudoelasticity of NiTi is currently overshadowed by the short fatigue life of NiTi wires used in this low cycle (200–2000 rpm), high amplitude ($\varepsilon_a > 2.5\%$) application, resulting in *in vivo* fracture or premature retirement of otherwise reusable NiTi-based wire devices. In this study, the failure of pseudoelastic 55.8 wt % Ni-Ti wire is investigated experimentally, as a function of experimental parameters that include the clinically relevant regime. The effects of radius of curvature, angle of curvature, wire diameter, strain amplitude, cyclic frequency, volume under strain, and specific heat of the surrounding environmental fluid are considered systematically. These data indicate that the lifetime or cycles to failure N_f of a rotating NiTi wire can be predicted via a modified Coffin-Manson relation that is a strong function of both strain amplitude and volume under strain, and a weaker function of frequency and fluid specific heat. The resulting quantitative relation can be used to predict useful device lifetime under clinically relevant conditions and thereby reduce incidences of *in vivo* failure. © 2004 Wiley Periodicals, Inc. J Biomed Mater Res Part B: Appl Biomater 72B: 17–26, 2005

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INTRODUCTION

NiTi is a shape memory alloy (SMA) that can be processed to exhibit both shape memory (SM) and pseudoelasticity (PE). These properties are due to the austenite–martensite transformation that is induced as a function of stress (PE) and temperature (SM). Pseudoelasticity confers reversible strains ranging from 8[1–4] to 15%,[5] depending on composition and processing history. This characteristic has made NiTi a useful biomedical material in both medical and dental applications that require insertion into or around narrow, curved canals. In the dental specialty of endodontics, NiTi wires, or files, are rotated under low-cycle/high-amplitude conditions to enlarge the root canal space. In fact, Walia et al.[6] introduced this material to the endodontic community in the 1980s, showing that the comparably greater elastic compliance of Nitinol™, (3M Unitek, MN) compared to stainless steel allowed improved access to curved root canals. The anatomy encountered in these root canals is highly variable,[7] and can subject the wire to severe curvatures and strain amplitudes in excess of 15%. Importantly, the catastrophic failure of NiTi files in endodontic practice is also highly variable, leading to compromised prognoses when these instruments fail *in vivo*—necessitating additional corrective surgery—and to elevated procedure costs when used files are retired prematurely.[8] In fact, although fatigue conditions vary and techniques differ among clinicians, typically a NiTi file is cycled in the canal for a total of 300 s at a frequency of 300 rpm or 5 Hz, and is retired from use and/or has failed catastrophically in less than 2000 cycles. The operating frequency recommended by manufacturers varies by type, and is typically 300 rpm for files fluted along the shaft and 2000 rpm for files fluted only at the distal end.

Although several investigations of low-cycle fatigue of pseudoelastic NiTi have been reported in the materials engineering[1,2,4,9–14] and endodontics communities,[15–17] the studies contributed by these distinct research communities cannot

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be compared to predict NiTi failure under general fatigue conditions. This shortcoming is due chiefly to the distinct approaches to research adopted by these two fields. Materials engineering studies such as those reported by Tobushi et al.[11–14] relate the cycles to failure N_f of pseudoelastic NiTi as a function of systematically varied and normalized quantities such as strain amplitude and relative temperature. In such studies, experiments are conducted on model NiTi specimens—smooth wires for which the cross-sectional diameter and the strain amplitude are constant over the length of the wire—but do not consider the range of strain amplitudes consistent with endodontic applications. Endodontic studies naturally consider operating conditions typical of the clinical setting, but often report lifetime as a comparison among actual instruments from different manufacturers, subjected to the same conditions. Even when instrument geometry and composition are maintained constant within a given study, operating parameters are not reported quantitatively in terms of strain, but rather as a function of curve radius and/or angle, from which strain amplitude can only be approximated. Quantitative comparisons among available endodontic studies are also inhibited by the various specimen geometries implemented, including fluted or tapered files that exhibit stress concentrations, position-dependent strain amplitude, and variable torque when in contact with the canal. Moreover, neither research community has considered whether and how the volume of material under strain impacts fatigue life.

In the present study, a novel rotating-bending fatigue test apparatus was designed to investigate systematically the influences of strain amplitude ε_a; frequency as revolutions per minute or rpm, specific heat of the surrounding fluid environment $c_{p,e}$; and volume of material under strain V_ε on the fatigue life of NiTi wires. Such an experimental investigation was employed to provide predictions of NiTi wire device life under the operating conditions typical of endodontic files, as well as to characterize NiTi fatigue and failure at large strain amplitudes. It is important to note that, to quantify these effects accurately in the present study, additional factors such as geometry-induced stress concentrations and torque were neglected and/or minimized.

MATERIALS AND METHODS

Materials

All samples used in the current study were cold-drawn from Type 508 55.8 wt % Ni-Ti wire of uniform diameter. Trace elements in this alloy include O, C, and Fe, each at approximately 0.05 wt %.[5]

Files were donated for this study by Lightspeed Technologies, San Antonio, TX. These files differ from commercially available Lightspeed files in that no cutting tip or fluting is present, and wires are drawn rather than machined to achieve specific file diameters.

Although the exact thermomechanical processing history of these wires is proprietary and strongly affects the

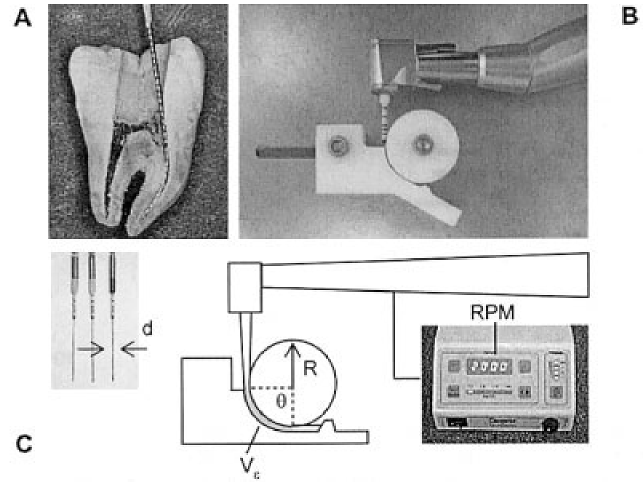

Figure 1. (A) Cross-section indicating canal curvature typical of endodontic applications. (B) Experimental rig designed for this study. Separate Teflon fixtures denote the canal radius (cylinder) and volume under strain (angled mount bracket) for a file rotated via a commercially available geared handpiece. (C) 55.8 wt % Ni-Ti files of 0.20; 0.32; and 0.45-mm uniform diameter were employed, and experimental parameters were defined as illustrated, where θ defines the fraction of the wire length and volume V_ε in contact with the cylinder of radius *R,* and Rpm or frequency *f* is maintained via a commercially available motor/power supply.

pseudoelastic properties, typical processing of pseudoelastic NiTi-based wires includes vacuum casting of an ingot; hot forging, rolling, and drawing at $T \sim 1000$ K to reduce ingot diameter; cold drawing at a low rate (10% area reduction/pass) to achieve final diameter; oxide removal through mechanical grinding or acid pickling; and annealing ($T = 400$–600 K) under tension to straighten.[18,19] For cold-drawn pseudoelastic wires of Ni content >55.5%, it is typical that heat treatment includes solution annealing at $T = 900$–1500 K followed by aging at $T = 700$ K, resulting in precipitation of Ni-rich phases.[19]

Wire diameters of 0.20, 0.32, and 0.45 mm, typical of endodontic applications, were tested. NiTi wire length was 31 mm for all diameters, with an additional 15 mm brass latch-type adaptor attached by the manufacturer that enabled insertion and rotation via a range of commercially available gear-reducing handpieces. This specimen geometry was chosen to facilitate both the full quantification of the stress and strain states and the clinical operating conditions (localized deformation and narrow frequency range).

Experimental Apparatus Design

An example of the experimental apparatus is shown in Figure 1. Teflon components were machined to maintain the NiTi wire curvature over well-defined volumes and under minimal wear-induced torque. Cylinders of 2-, 4-, and 8-mm radii defined the radius of curvature of the canal *R.* For each *R,* three mount brackets were designed to hold differing fractions of wire length against the cylinder, allowing the volume under strain V_ε to be varied systematically. A commercially

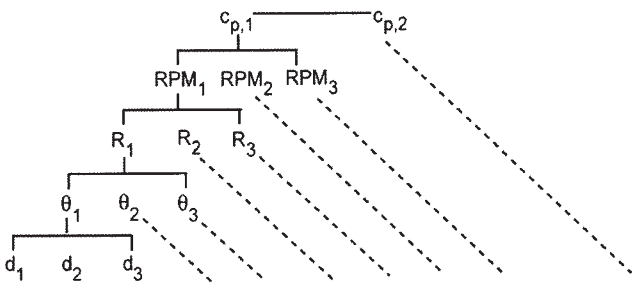

Figure 2. Matrix of experimental conditions. For a given environmental fluid of known specific heat C_p, frequency *f*, canal radius *R*, canal angle θ, and file diameter *d* were varied systematically. This resulted in 81 parameter sets in each of two environments, and facilitated comparisons among groups for fixed values of these variables.

available dental electric motor (Aseptico, Woodinville, WA.) was used to rotate the wires at constant speed.

This apparatus facilitated three specific values of V_ε to be maintained for each of three specific values of *R,* for each of three file diameters *d.* Maximum strain amplitude ε_a is defined by both the file diameter *d* and the canal radius *R:*

$$\varepsilon_a = 2d/(2R + d) \tag{1}$$

and is constant over the surface area covering the volume of wire material defined by the mount bracket for a given *R:*

$$V_\varepsilon = ([\pi^2 d^2 R]/2) \cdot (\theta/360°) \tag{2}$$

where $\theta/360°$ is the angular fraction over which the wire contacts the cylinder of radius *R.*[20] Note that the strain amplitude actually increases linearly from the neutral axis (approximately center) of the file to the outermost surface, such that the surface of the file under maximum extension is under maximum tensile strain while the opposite surface of the file is under (an equal and opposite magnitude of) maximum compressive strain.

Fatigue Testing

A matrix of 81 unique fatigue parameter sets were conducted in each of two fluid environments. As shown in Figure 2, strain amplitude ε_a, volume under strain V_ε, revolutions per minute rpm, and the specific heat of the surrounding fluid environment $c_{p,e}$ were varied systematically, and the time to failure was monitored. The strain amplitude ε_a ranged from 2.5 to 18.2%. Rotational frequency was maintained at 200, 950, or 2000 rpm, equivalent to frequencies of 3, 16, and 33 Hz and consistent with the range of frequencies employed for commercially manufactured NiTi files in endodontic applications. The specific heat of the surrounding environmental fluid normalized by that of the wire material $c_{p,e}/c_{p,NiTi}$ was 1.2 and 5.0 for air and water environments, respectively. For each parameter set, four experiments were conducted in air ($T = 300$ K), and two in water ($T = 300$ K), for a total of 486 experiments.

RESULTS AND DISCUSSION

The relationship between ε_a and N_f obtained by this investigation is shown in Figure 3, along with the results of similar investigations reported in the literature. As evidenced by Figure 3, Tobushi et al.[11–14] have contributed significantly to the study of NiTi rotating-bending fatigue for ε_a limited to 0.5–2.5%. These materials engineering studies utilized NiTi wire of slightly larger diameter (0.75 mm), and systematically varied temperature, strain amplitude, environmental fluid, and rotational speed (100–1000 rpm). Pruett et al.[17] contributed the first NiTi fatigue experiments in the endodontic literature. NiTi wire of differing shaft diameters were utilized, and rotational speed (750–2000 rpm) exceeding that of the Tobushi et al. studies was monitored. Here, the angle and radius of curvature were recorded and varied, but N_f was evaluated as a function of file size rather than as a function of quantified strain amplitude over a defined material volume. Below, we discuss the results of the current study as a function of each of the systematically varied parameters, and in comparison to the trends shown in Figure 3.

Effect of Strain Amplitude on Cycles to Failure

The effect of ε_a on N_f in air is shown in Figure 4. This logarithmic relationship is representative of classic low-cycle fatigue, which is governed by accumulated plastic strain and can be expressed by the Coffin-Manson low-cycle fatigue relation as follows:

$$\varepsilon_a N_f^\beta = \alpha \tag{3}$$

which can be expressed explicitly as a prediction of cycles to failure N_f as

$$N_f = [\alpha/\varepsilon_a]^{1/\beta} \tag{4}$$

where the moduli α and β are considered empirical material constants.[21] Below, we consider these fatigue life moduli α and β as functions of additional operational and environmental parameters such as rpm and $c_{p,e}/c_{p,NiTi}$.

Effect of RPM on Cycles to Failure

The relationship between ε_a and N_f at varying rpm in air is shown in Figure 5 (A). Although N_f decreases as rpm increases, this relationship is not as straightforward as the logarithmic dependence of N_f on ε_a. In fact, these data show that the effect of frequency on N_f is best expressed as logarithmic functions of frequency, α and β:

$$\alpha = -0.76 \ln[\text{rpm}] + \kappa_1 \tag{5}$$

$$\beta = \kappa_2 \ln[\text{rpm}] - \kappa_3 \tag{6}$$

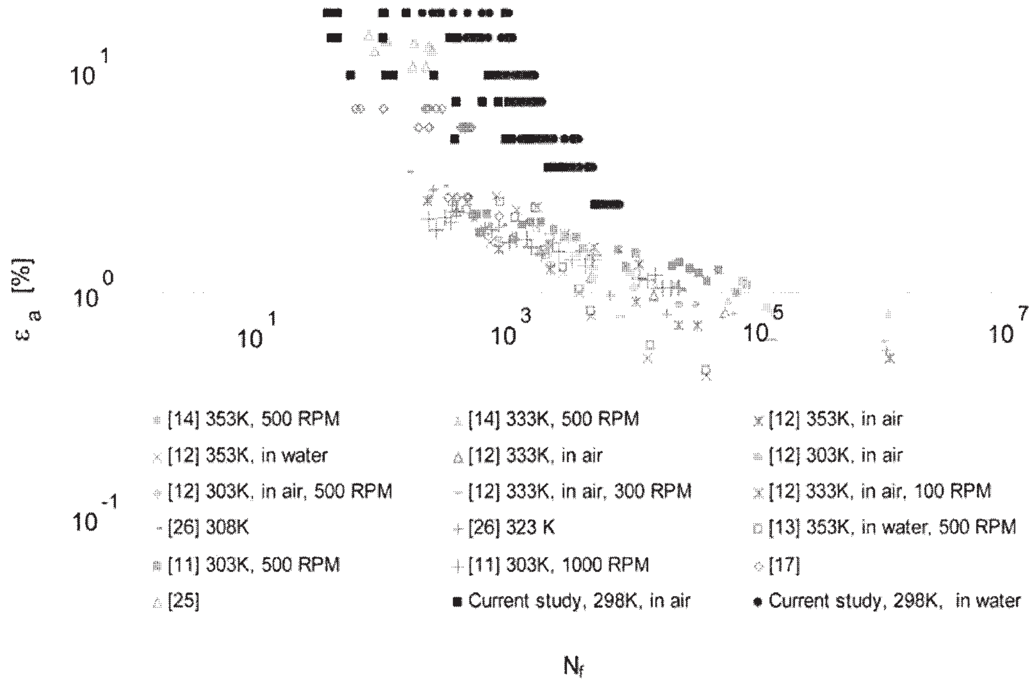

Figure 3. Comparison of data in the current study with that reported to date in the literature on NiTi SMA wire under rotating-bending fatigue. At large strain amplitudes, data in the current study are consistent with that reported in the endodontics literature. At strain amplitudes < 1%, N_f of the current study exceed that obtained in the materials engineering literature.

where κ_i are identified below as functions of environmental conditions. This logarithmic dependence of the fatigue moduli was determined empirically, and does not derive from an established physical relationship among frequency, environment and fatigue life. Figure 5(B) and 5(C) demonstrate the logarithmic dependence of α and β on rpm. This nonlinear relationship rationalizes a discrepancy found in the literature between endodontic and materials engineering investigations. In the endodontic research community, Pruett et al.[17] reported a negligible effect of rpm on N_f for $\varepsilon_a > 2.5\%$ and rpm > 1000. Conversely, the materials engineering analysis of Tobushi et al.[11] indicated a strong effect of rpm on N_f for $\varepsilon_a < 2.5\%$ and rpm < 1000. The logarithmic relationship quantified by the expanded range of experimental conditions in the current study explains this apparent contradiction well. That is, rotational speed of NiTi wire does impact fatigue life over the range of clinically relevant ε_a and rpm, but this effect is greatest for small ε_a and low rpm.

Effect of Relative Specific Heat on N_f

In separate studies, Tobushi et al.[14] have shown that the temperature of NiTi wire increases under low-cycle, high-amplitude fatigue conditions, caused ostensibly by the martensite–austenite phase transformation over each cycle. Certainly, the effect of NiTi temperature on N_f could be measured and expressed explicitly in the present study. However, as the purpose of the current study is to predict NiTi file lifetime as a function of parameters that can be quantified or controlled in a clinical setting, we have purposely considered lifetime as a function of the properties of the fluid environment in which NiTi file fatigue occurs.

The effect of relative specific heat on N_f is also shown in Figure 6. It is evident that N_f increases and lifetime variability decreases with increasing $c_{p,e}/c_{p,NiTi}$. Physically, this means that the capacity for the surrounding fluid to absorb the heat generated by the cycling NiTi wire correlates with increased lifetime, and with increased predictability of that lifetime. The specific heat of NiTi is 0.20 cal/g°C.[3] Thus, when NiTi wire is cycled in an environmental fluid such as water (c_p = 1.00 cal/g°C), the fluid acts as an efficient heat sink and absorbs the energy generated by the transforming NiTi without increasing appreciably in temperature. In contrast, when NiTi wire is cycled in an environmental fluid such as air (c_p = 0.24 cal/g°C),[22] the fluid effectively insulates the rotating NiTi file, inhibiting heat dissipation and thus decreasing N_f. This is illustrated clearly in Figure 5(B) and 5(C) through the differential effects on α and β in air and in water. For the range of ε_a and rpm considered in the present study, the moduli α and β vary linearly with relative specific heat as:

$$\alpha = -0.76 \ln[\mathrm{rpm}] + 2.10[c_{p,e}/c_{p,NiTi}] \tag{7}$$

$$\beta = 0.10[c_{p,e}/c_{p,NiTi}] \ln[\mathrm{rpm}] - 0.75 \tag{8}$$

From these empirical relationships, it is clear that the capacity to which the fluid surrounding the rotating NiTi file can

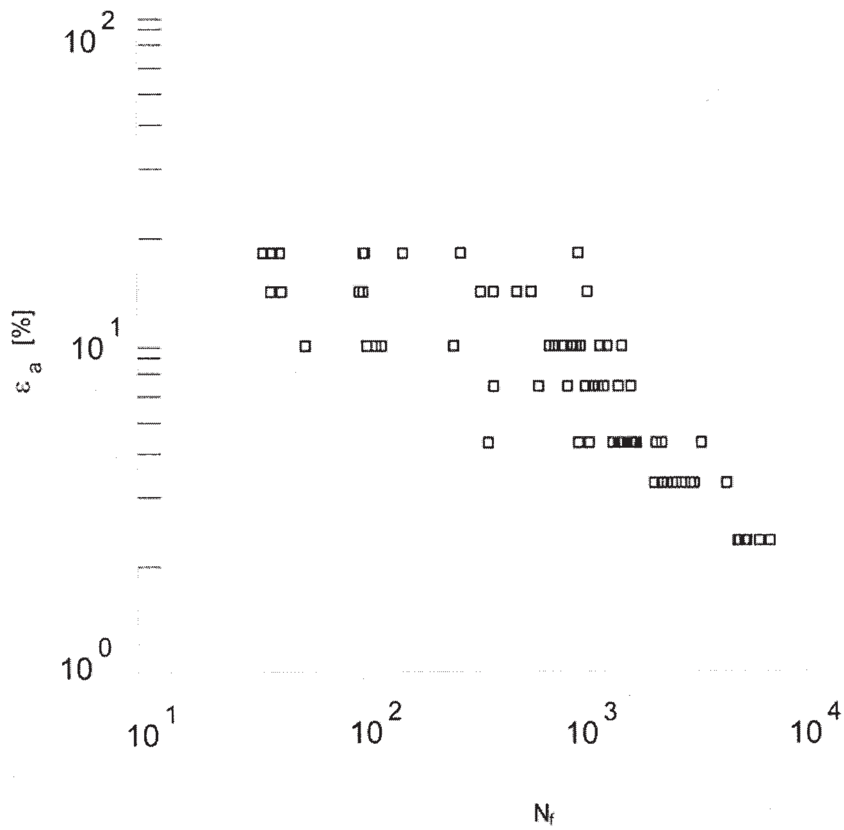

Figure 4. Cycles to failure N_f as a function of strain amplitude ε_a of NiTi SMA wire under rotating-bending fatigue in air. These data represent a larger range and larger strain amplitudes than reported previously, and include the range of clinically relevant parameters (including ε_a up to 18%, typical of endodontic applications) and environments.

absorb the heat generated by the cyclic phase transformation impacts device lifetime. For this reason, a thermal parameter such as $c_{p,e}/c_{p,NiTi}$ or relative thermal diffusivity should be considered carefully in future studies and in clinical applications. Note that increased specimen temperature is typical of metals under fatigue, and is not a unique characteristic of NiTi fatigue. However, in metals and alloys that do not undergo stress-induced phase transformations, this temperature increase is attributed to dislocation activity, rather than to crystal structure transformations. In the current study, significant dislocation activity may occur for $\varepsilon_a > 8\%$, as this magnitude exceeds the lower limit[1–4] of reversible strain for NiTi-based alloys. Note that the magnitude of the reversible strain limit is dependent on composition and thermomechanical processing.

Effect of Strain Volume on N_f

The effect of the volume under strain V_ε on N_f at varying ε_a is shown in Figure 7. Over the range of fatigue parameters considered, N_f is a strong function of the volume of NiTi under strain, or strain volume. In general, lifetime varies inversely with V_ε, i.e., N_f decreases linearly with increasing V_ε [Fig. 7(A)]. However, analysis of specific parameter sets indicates that the effects of strain amplitude and strain volume are in fact coupled. Figure 7(B) shows results for a fixed file diameter of 0.2 mm, where the same magnitudes of ε_a are applied at three different rpms. Figure 7(C) indicates the strength of the volume effect as a function of ε_a, where strength is quantified as the slope of the N_f versus V_ε data in Figure 7B, $\Delta N_f/\Delta V_\varepsilon$. As the absolute value of the slope in Figure 7(C) increases, the strength of the volume effect increases. That is, N_f decreases more sharply with increasing V_ε for increasing $\Delta N_f/\Delta V_\varepsilon$. Clearly, the effect of strain volume is not independent of strain amplitude, but rather the strain volume effect is most pronounced at the extreme strain amplitudes considered.

As the fatigue data follow an empirical relationship based on accumulated strain,[21] it is plausible to rationalize the strain volume effect as follows: when strain amplitude is low, the defect volume generated per cycle is also small with respect to the strain volume. In this context, the strain volume fills incrementally with defect volume, and the defect volume is surrounded by relatively defect-free material that transformed reversibly under the same strain. If we idealize each defect volume as a slice within the strain volume, the number of slices that can fit within the strain volume will correlate closely with the number of cycles to failure and thus lifetime would increase with increasing strain volume. However, these data show that N_f actually

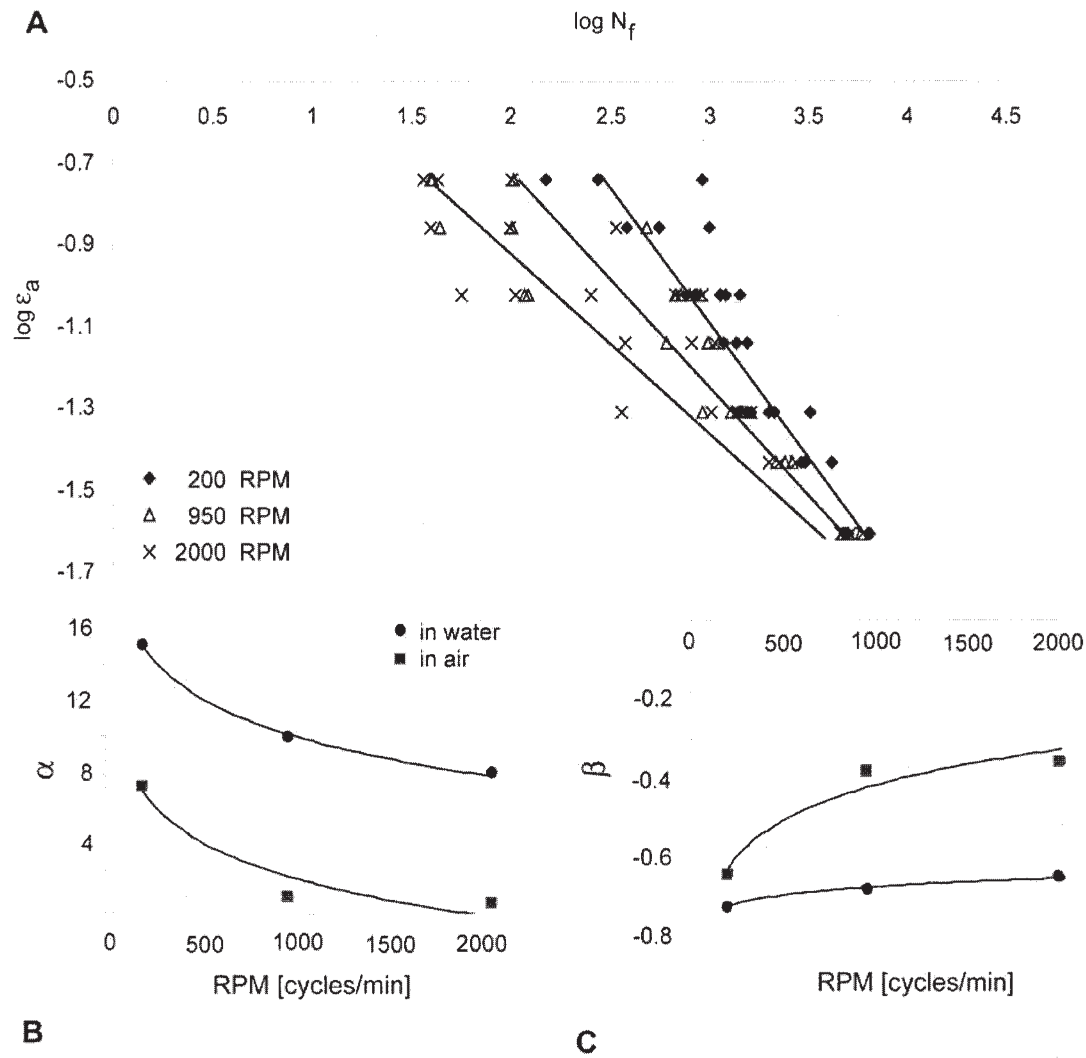

Figure 5. (A) Effect of cyclic frequency rpm on N_f. (B) α, and (C) β are moduli of N_f in the Coffin-Manson relation and depend logarithmically on rpm. α decreases with increasing rpm, and β decreases with increasing rpm.

decreases with increasing V_ε for all parameters considered. That is, the greater the volume of material under strain, the shorter the lifetime. This finding suggests that the transformed material and/or defects resulting from incomplete transformation[23] or plastic deformation of the martensitic phase interface immediately with the unstrained material volume that constrains V_ε. Thus, it appears that the strain volume directly limits the lifetime by increasing the number of defects that can be accommodated within V_ε and subsequently contribute to defect generation at the strained/unstrained interface.

Here, it is important to note that the peak effect of strain volume in Figure 7(C), indicating the coupled effect of V_ε and ε_a on N_f, was also observed for file diameters of 0.32 and 0.45 mm at certain rpms, so this is not a result unique to a particular file geometry. However, this trend was not observed for the entire range of rpms. For example, the strength of the volume effect increased linearly with increasing ε_a for a file diameter of 0.32 mm at 200 rpm (3 Hz). Thus, this is a strong but not universal trend that may also depend on whether the magnitude of ε_a exceeds the reversible strain limit. The peak effect may indicate a transition in defect microstructure from martensitic variants (at $\varepsilon_a <$ transformation strain) and plastic deformation of residual martensite (at $\varepsilon_a >$ transformation strain), but this has not been confirmed through postfracture analysis. We are currently attempting to quantify the strain volume effect more closely, but the data discussed herein show simply that strain volume varies inversely with N_f and is therefore an important parameter in the characterization of the rotating, bending failure of NiTi. Further, these data support a damage accumulation model that—although unexpected for the pseudoelastic phase transformation of NiTi at strain amplitudes less than 15%—explains well the effects of these operating parameters on NiTi file lifetime.

Including this inverse effect of strain volume V_ε and, for the present, neglecting the coupled effect of strain amplitude on this effect, we arrive at the following empirical expression

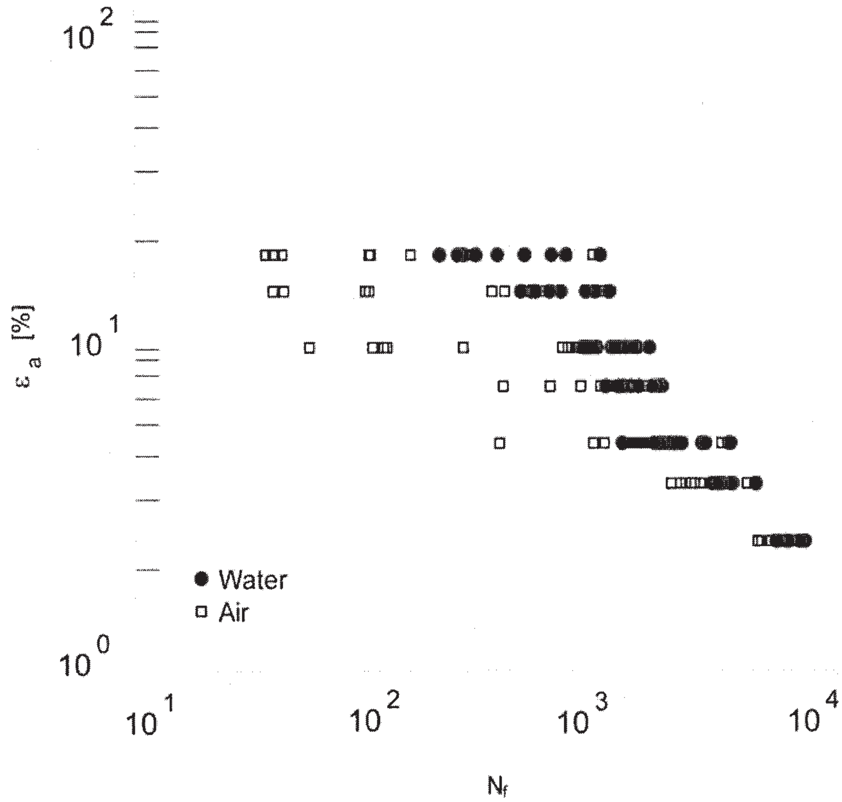

Figure 6. Comparison of fatigue lifetime as a function of surrounding environmental fluid, over identical ranges of strain amplitude, strain volume and RPM. Lifetime increases and variability in lifetime decreases in water with respect to in air.

that predicts NiTi-based device lifetime for a clinically relevant range of experimental conditions:

$$N_f = V_\varepsilon^{-1}[(-0.76 \ln[\text{rpm}] + 2.10[c'])/\varepsilon_a]^{1/(0.10c' \ln[\text{rpm}] - 0.75)} \tag{9}$$

where Eq. 4 is now expressed by empirically determined functions of operating and environmental conditions, and c' is the ratio of environmental fluid to NiTi specific heat $c_{p,e}/c_{p,NiTi}$.

Fracture Surface Analysis

Subsets of the experimental matrix were investigated via scanning electron microscopy [JEOL JSM-5510LV (Peabody, MA), in secondary electron mode] to examine the fracture surface. Figure 8 shows the fracture surfaces as a function of ε_a, for a fixed cyclic frequency of 200 rpm (3 Hz) and a fixed strain volume of 0.26 mm^3. Note that the fracture mode is primarily ductile for all levels of strain amplitude, as indicated by the dimpled cup-and-cone fracture surface,[21] and that striations typical of accumulated fatigue damage are not observed at any strain amplitude. In fact, striations were not observed on any fracture surface analyzed in this manner. Internal voids were also observed, and the ductile surface features were on the order of the grain size (15 μm). Gross, residual plastic strain in the form of permanent deformation of the file geometry was observed in samples for which $\varepsilon_a > 15\%$ [e.g., Fig. 8(E)], but the fracture surface did not reflect failure initiation from the tensile surface of the permanently deformed file. The composition of the material surrounding the metallic file was identified through energy dispersion spectroscopy (EDS) as a F- and C-rich coating that is not present on the as-received files (data not shown). As Teflon is a fluoropolymer,[24] this coating is thus identified as material resulting from minimal wear against the Teflon fatigue apparatus.

Comparison with Previous Results and Clinical Relevance

Figure 3 compares the results of this study with those reported in the literature by both the materials engineering and endodontics research communities, including those studies not discussed explicitly in the text.[25,26] At large strain amplitudes ($\varepsilon_a > 5\%$) in air environments, the current study is consistent with the findings of Pruett et al.[17] and Dietz et al.,[25] two studies from the endodontics community that considered large file deformations but did not report quantitative measures of strain. Here, we have calculated ε_a from the available data provided in the articles, according to Eq. 1. At smaller strain amplitudes, the device lifetime measured in the current study significantly exceeds that measured by others, either in air or in water environments. For

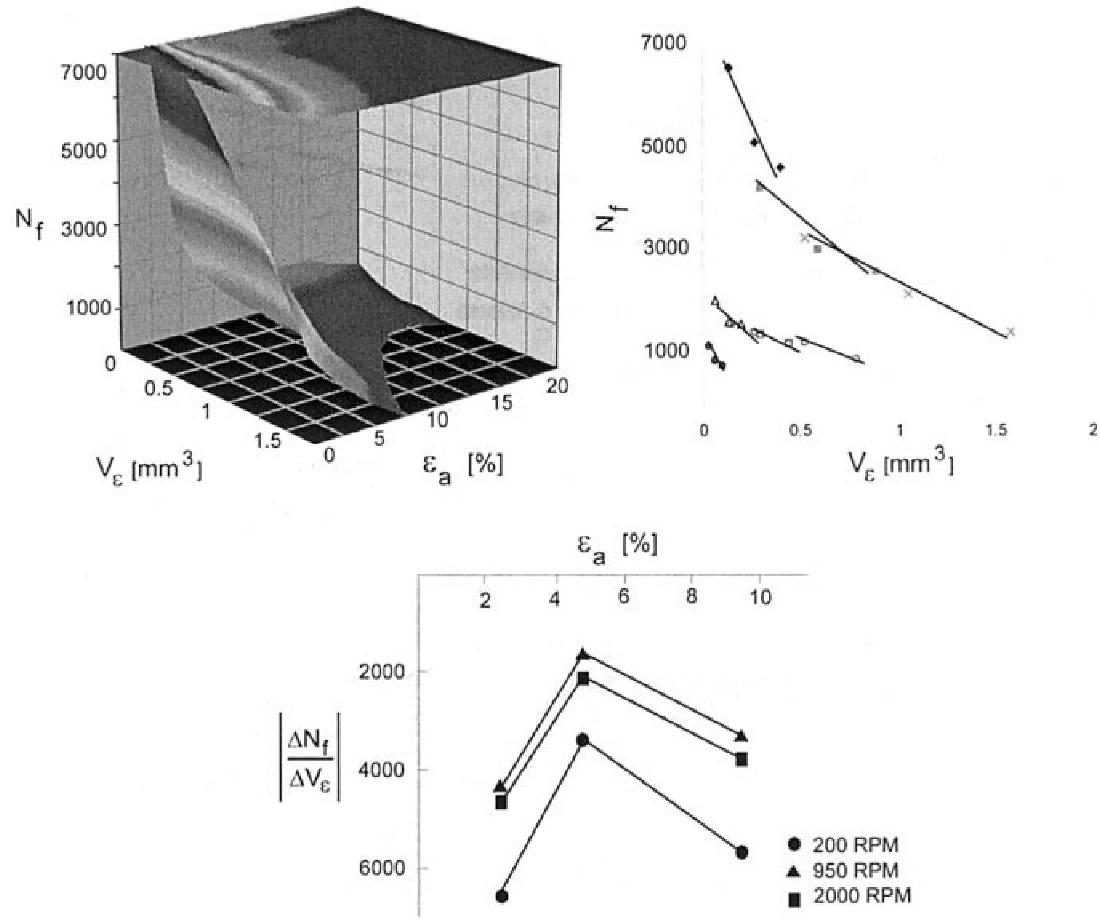

Figure 7. (A) N_f is a strong function of both ε_a and V_ε. (B) N_f decreases linearly with increasing V_ε. Here, each symbol series represents a fixed value of strain amplitude at 200 rpm. (C) For a fixed diameter and strain amplitude, the strength of the volume effect, represented as the slope of the data in that series from (B), is maximal at the extreme strain amplitudes considered.

example, at a strain amplitude $\varepsilon_a \sim 2.3\%$ in air, average lifetime was measured as 530 cycles and 372 cycles in the studies of Pruett et al.[17] and Tobushi et al.,[11] respectively, whereas $N_f =$ 4663 cycles in the current study.

There are several possible explanations for this discrepancy. First and foremost, this contrast underlines the clear dependence of NiTi device lifetime on variables other than cyclic strain amplitude ε_a. In the example cited above, Tobushi et al. operated at frequencies (1000 rpm) comparable to the current study, but subjected the entire file volume to uniform strain, i.e., $V_\varepsilon/V_{total} =$ 100%. Second, as is consistent with our hypothesis of accumulated damage from incomplete phase transformation and/or plastic deformation of the transformed phase, composition and processing history will affect critically the stress required to induce plastic deformation and also the density of preexisting defects within the material. The NiTi-based material used in the current study was cold-drawn without indication of subsequent machining, whereas the NiTi-based files used by Pruett et al.[17] were mechanically ground to attain specified file diameters and distal fluting. Even if the same chemical composition were utilized in both studies, the surface machining of the files used in studies such as Pruett et al. results in a higher density of defects such as surface microcracks that may act as failure initiation sites in service, even after subsequent annealing treatments, and also affects the strain required to induce phase transformation and plastic deformation.[19]

Although the NiTi device lifetime predicted empirically by Eq. 9 relies on several independent variables, this study was framed in the context of *material* behavior such that these results are immediately relevant to a clinical setting and can be made readily accessible to the end user. NiTi file dimensions are well-quantified, and three-dimensional geometry of the canal in which such devices are used can be quantified through approaches such as microcomputed tomography (microCT).[27, 28] Together with the operating parameters of the device, these data could be entered into an automated program that would inform the clinician of the recommended useable lifetime of that particular file under the specific environmental and anatomical conditions.

The dependence of N_f on rpm, $c_{p,e}/c_{p,NiTi}$ and V_ε is consistent with the concept that the fatigue life of NiTi is limited by microstructural damage accumulated per cycle within a finite volume. In fact, there was no measureable change in lifetime trends when the strain amplitude exceeded the range

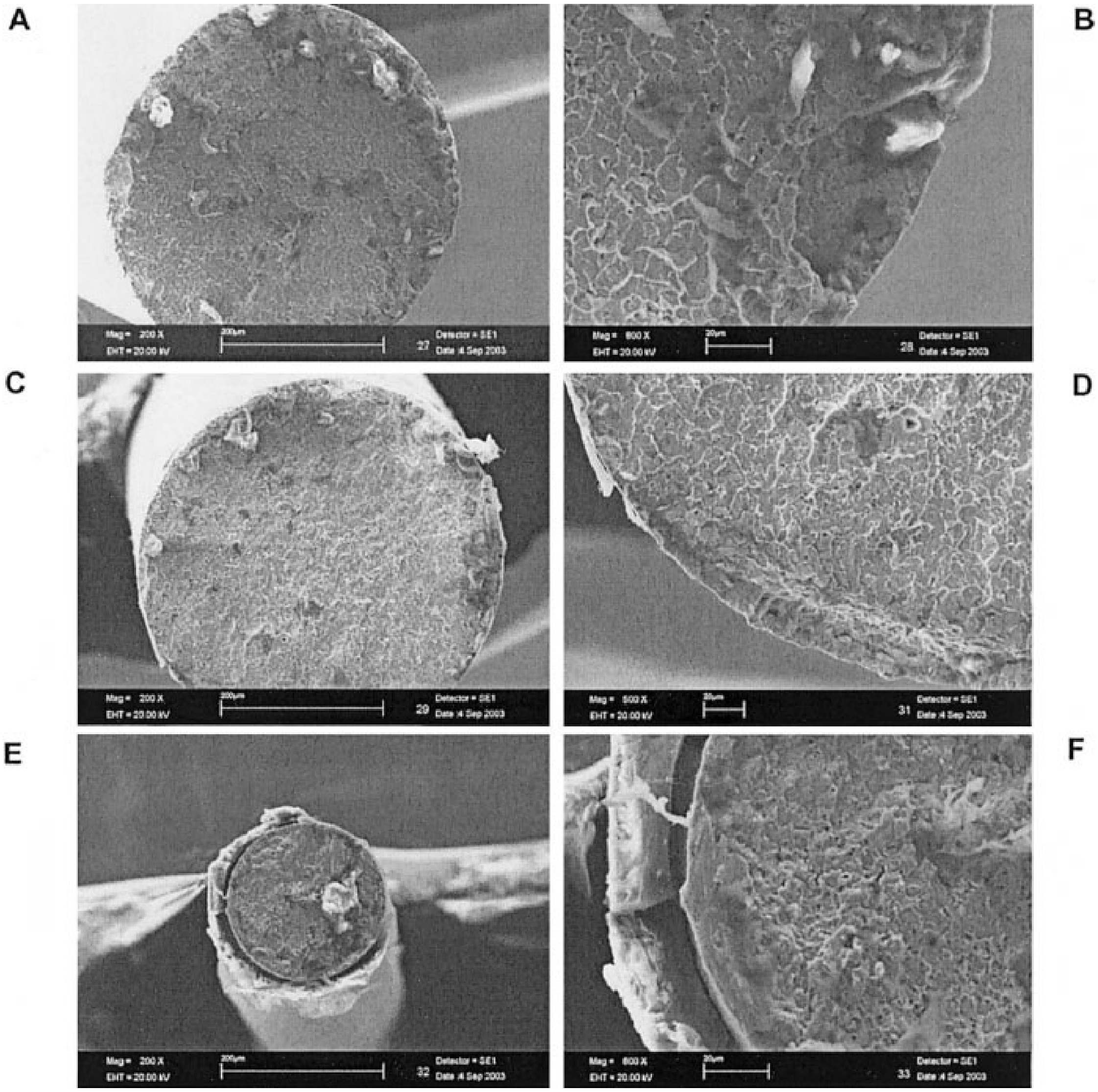

Figure 8. Scanning electron micrographs of fracture surfaces for fixed rpm = 200 and $V_\varepsilon = 0.26$ mm^3. (A) and (B), $\varepsilon_a = 2.5\%$, well within the reversible strain limit of NiTi. (C) and (D), $\varepsilon_a = 9.5\%$, approaching irreversible strain; (E) and (F) $\varepsilon_a = 18.2\%$, beyond the reversible strain limit. Note that all fracture surfaces are primarily ductile, that no striations are observed, and that fracture initiation site is not immediately obvious. Scalebars are 200 μm in (A), (C) and (E); 20 μm in (B), (D), and (F).

of phase transformation strains. Thus, these data indicate that the martensite–austenite transformation may not be fully reversible over the range of experimental conditions explored, even when the applied strain amplitude is less than the reversible strain limit under uniaxial fatigue. Another plausible explanation is that the transformation front may generate defects, martensitic variants, and residual martensite at the interface with unstrained material volumes. It has been shown that both primary defects (caused by thermomechanical processing) and secondary defects (caused by mechanical fatigue) limit fatigue life of pseudoelastic, equiatomic NiTi,[4] and the current study supports this finding for 55.8 wt % Ni-Ti over a wide range of low cycle-high amplitude conditions. As we continue to develop experiments and models that can explain further the scientific underpinnings of the surprisingly short lifetimes of NiTi under this fatigue environment, the above experimental results provide a range of strain amplitudes extended significantly beyond that considered previously in the engineering community. Further, this analysis provides a prediction of device lifetime that is relevant to clinical conditions, and is based on material behavior rather than device-specific parameters. Although the above study neglects the application and analysis of torque during rotating-bending fatigue that are particularly relevant to fluted NiTi endodontic file geometries, the data presented herein provide the clinician with device-independent information to reduce the incidence of *in vivo* fracture of NiTi files. Clearly, the implementation of an aqueous environment during instrumenting of the root canal will prolong file lifetime and decrease lifetime variability. In addition, minimizing file diameter, especially in canal geometries that induce large strain amplitudes, will provide the clinician with greater useful instrument lifetimes. Future studies aim to include the effect of thermal properties, torque, and complex cross-sectional geometries on NiTi file lifetime. A testing apparatus identical in design to the one used in this experiment, but formed from hydroxyapatite to mimic the surface roughness and mechanical properties of tooth dentin, is currently in development.

CONCLUSIONS

1. Lifetime of Ni-55.8 wt % Ti wires subjected to rotating bending fatigue is well-characterized by the Coffin-Manson model of accumulated strain. This indicates that, although temporal lifetimes of endodontic files can be increased through intermittent application of force, the number of cycles that a file can withstand prior to fracture is fixed by the total amount of strain accumulated in each cycle.
2. A fatigue equation is proposed incorporating the effects of ε_a, V_ε, rpm, and $c_{p,e}/c_{p,NiTi}$.
3. N_f is a strong function of ε_a and V_ε, and a relatively weaker function of rpm and $c_{p,e}$.
4. Strain volume V_ε is an important parameter in NiTi fatigue characterization. N_f varies inversely with V_ε, but in a manner dependent on ε_a.
5. An apparent inconsistency regarding the effect of rpm on N_f is rationalized through the range of experimental parameters as a logarithmic dependence in the proposed fatigue equation.
6. In clinical applications, file lifetime N_f is increased and predicted more accurately via reduction of the ratio of file diameter d to canal radius R, minimization of the volume of file under strain V_ε, and operation in fluid environments that act effectively as heat sinks during mechanical fatigue.

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