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In Vivo Force Decay of Niti Closed Coil Springs

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Abstract

Introduction—Nickel-titanium (NiTi) closed coil springs are purported to deliver constant forces over extended ranges of activation and working times. In vivo studies supporting this claim are limited. The objective of this study is to evaluate changes in force decay properties of NiTi closed coil springs after clinical use.

Methods—Pseudoelastic force-deflection curves for 30 NiTi coil springs (used intra-orally) and 15 matched laboratory control springs (simulated intra-oral conditions - artificial saliva, 37°C) were tested pre- and post-retrieval via Dynamic Mechanical Analysis (DMA) and the Instron machine, respectively, to evaluate amount of force loss and hysteresis change following 4, 8, or 12 weeks of working time (n=10 per group). Effect of the oral environment and clinical use on force properties were evaluated by comparing in vivo and in vitro data.

Results—The springs studied showed a statistically significant decrease in force (~12%) following 4 weeks of clinical use ($p<0.01$), with a further significant decrease (~7%) from 4–8 weeks ($p=0.03$) and force levels appearing to remain steady thereafter. Clinical space closure at an average rate of 0.91mm per month was still observed despite this decrease in force. In vivo and in vitro force loss data were not statistically different.

Conclusions—NiTi closed coil springs do not deliver constant forces when used intra-orally, but they still allow for space closure rates of ~1mm/month.

INTRODUCTION

Space closure is an important aspect of orthodontic treatment. Using light, continuous forces over a relatively long activation range (e.g., 5–10 mm space closure) allows for more biologically favorable and clinically efficient tooth movement with fewer negative side effects.[1–4] Some common orthodontic materials used for space closure deliver very high initial forces that decay rapidly prior to re-activation.[5–8] NiTi materials are purported to overcome this rapid force decay problem and supply light, continuous forces over a long activation range.[9,10] The claim that they are able to deliver these more biologically favorable forces, and thus potentially lead to more efficient tooth movement, is how many clinicians have justified their use despite their relatively greater cost compared to other common space closing materials such as elastomeric chain or closing loops.

NiTi alloys possess the unique properties of shape memory and superelasticity because of their capacity to alter their crystalline bonding patterns between the martensitic and austenitic phases as a function of temperature and applied stress without permanent dislocation of atoms.[11, 12] Thus, these materials exhibit a relatively flat (non-linear) load-deflection curve signifying the superelastic characteristic of NiTi in which it delivers a low, constant force over a long range of activation.

Unfortunately, several *in vitro* laboratory studies and limited *in vivo* studies have suggested that NiTi closed coil springs may not be delivering constant forces.[5, 13–17] Maganzini et al. examined 14 different types of NiTi closed coil springs *in vitro* from five different companies and concluded that most of the springs tested did not exhibit constant deactivation forces or physiologic peak load forces.[5]

It is known that *in vivo* clinical outcomes correlate poorly with *in vitro* studies. Intra-orally, plaque accumulation on appliances and the array of oral bacteria with their highly varied by-products creates a unique environment that is not well simulated in the laboratory.[18] Additionally, the mechanical properties of NiTi products have been shown to be highly dependent upon intra-oral temperature changes.[12, 18, 19] Eliades et al. highlighted the need for post-clinical retrieval analysis in order to fully understand the *in vivo* material properties.[18] Clinical data examining force loss in the NiTi coil springs is limited, but *in vivo* results from Nightingale and Jones demonstrated average 48% force loss values following 22 weeks of intra-oral use.[14] This current project is aimed at evaluating the intrinsic force decay properties of NiTi closed coil springs following clinical use with the hypothesis that they will experience force decay proportional to stretch duration and thus affect efficacy of space closure. Comparisons with matched laboratory analysis will elucidate the overall effect of the intra-oral environment on spring properties.

Understanding the force decay properties of NiTi closed coil springs during intra-oral use will enable the clinician to make more cost effective decisions and empower them to deliver more efficient and effective treatment.

MATERIALS AND METHODS

Subjects

Patients in active treatment at the University of North Carolina School of Dentistry graduate orthodontic clinic or dental faculty practice that met the inclusion/exclusion criteria outlined below were eligible to participate in this study. Following IRB approval (study #10-1802) from the Institutional Review Board of the University of North Carolina, 11 patients were consecutively enrolled and data collected based on the following criteria:

<u>Inclusion Criteria</u>

- Have space closure treatment need
- Space closure performed with sliding mechanics using 150 gm GAC-Sentalloy NiTi coil springs
- Consent to participate in the study

<u>Exclusion Criteria</u>

- Springs that showed permanent deformation prior to or at the removal stage

Springs

All springs used in this study were GAC Sentalloy closed coil springs of medium grade which were advertised by the manufacturer to deliver a force of 150gm without deformation or force change when stretched in a range of 3 to 15mm.[9] These commercially available springs are approximately 9mm long (eyelet to eyelet) and consist of a 3mm length of coil with eyelets at each end. All springs used were from the same lot number (Lot No. B3X0). A segment of coil spring was cut and used to run differential scanning calorimetric (DSC, Q100, TA Instruments, New Castle, DE) analysis to investigate the phase transformation of the coil NiTi. The rate of temperature change was 1°C per minutes.

Mechanical Testing using Dynamic Mechanical Analysis (DMA) and Instron

Pre-Testing—Initial (pre-use) force levels for each spring were tested using Dynamic Mechanical Analysis (DMA). DMA is a technique that incorporates stress/ strain force measurements to study the mechanical properties of a material. DMA was chosen for this study due to its ability to 1) accurately control temperature (± 0.1°C) during force analysis and 2) perform a load-controlled test. Ten separate springs were stretched during preliminary testing and pseudoelastic force-deflection curves were generated for each. While there were small variations in force levels from each spring, the ideal testing force for DMA was determined to be 300 gm. At this force level, many of the preliminary springs were able to achieve almost the full 12mm activation range recommended by the manufacturer while beyond this force level, many springs were stretched beyond the 12mm guideline and possibly distorted. While the manufacturer's reported force level was 150grams, some of the springs were reaching 300grams at the stretch distance of clinical space closure. Testing the springs at 300grams allowed us to capture/characterize the entire range of loading and unloading force curves for these NiTi coil springs. This amount of force did not introduce plastic deformation according to our force-deflection plots.

All 55 springs used in this study were tested out on a DMA apparatus (Model, 2980, TA Instruments, New Castle, Delaware) at the constant force 300gm (Figure 1A). Temperature was controlled using a combination of liquid nitrogen and thermal heating to maintain temperature at 37°C +/− 0.1°C. Force was ramped at a rate of 0.5N/min up to 2.942N (300gm) and then back down to zero at the same rate. Each spring was pre-loaded to 0.24N (24gm) and maintained at the isothermal temperature of 37°C for 2 minutes prior to force ramping. Springs were attached to 0.032" SS hooks, which were gripped by the DMA film tension clamp (Figure 1B).

Post-Testing—The final (post-use) testing of the springs was conducted on an Instron universal testing machine (Model 4411, Norwood, Massachusetts) (Figure 1C), which uses an uniaxial load cell to measure forces and activations and generate load-deflection curves. Instron used displacement control rather than force control, which stretched each individual spring to the same length it was activated in pre-testing via DMA. The springs were stretched to this length at a rate of 5mm/min and then allowed to return to their original relaxed state at the same rate. The Instron was equipped with a +/−500 N static load cell (Instron, Serial Number UK 27) attached to the crosshead. A 0.032" SS hook was attached to the load cell to hold the upper loop of the coil spring. The opposite end of the coil spring was held by a 0.032" SS rod, which was inserted within a special testing jig that was attached to the instrument base. The spring and the special jig were contained within a double jacketed water chamber and submerged in water (Figure 1D). The temperature of the water within the inner bath (37° +/− 1°C) was maintained by circulating temperature controlled water through the outer closed chamber. This temperature was carefully regulated by a water circulating temperature controller (Haake, Germany). The monitoring thermocouple for the water circulator was placed within the inner chamber. The springs were maintained in the water bath for 2 minutes prior to stretching.

Groups

Following DMA pre-testing, the springs were randomly distributed to 3 different groups: 1) Clinical, 2) Laboratory, 3) Control.

1. Clinical springs (n=30): Thirty of the pre-tested springs were used during the treatment of patients in the orthodontic clinics at the University of North Carolina at Chapel Hill, from May 2011–December 2011. Once a patient met the above inclusion criteria, they were enrolled in the study and initial data regarding space closure was gathered. The springs were attached from the canine hook to the molar hook and ligated to the canine hook to prevent loss (Figure 2). Intra-oral measurements were made for spring activation range and interdental spacing using a Boley gauge. In addition, bracket slot size, wire size, and wire material were recorded. Patients were seen on their normal recall schedule of 4 weeks. At each recall, the springs were checked for signs of obvious distortion and maintained undisturbed until time of collection if no distortion was noted. Ten springs were collected for each of the time points in the study (4, 8, and 12 weeks). Upon removal of the springs, final measurements regarding spring activation length and remaining space were recorded. The springs were cleaned in a 10% Formalin solution for 10 minutes, rinsed with de-ionized water and stored dry in plastic bags for final force testing on the Instron.

Depending upon space closure needs of the patient, up to 4 springs could be collected from the same patient at a given time (one per quadrant). If additional space closure was needed following retrieval of the springs, new springs could have been placed in the same patient and collected after another 4, 8, or 12 weeks. Overall, 11 patients (7 Female, 4 male) participated in this prospective study (Average age 23.2 years, range 13–43 years). The slot size and type of archwire used for each clinical spring is summarized in Table 1.

2. Laboratory springs (n=15): Fifteen of the pre-tested springs were placed on SS plates with SS attachment pegs set at a distance of 20mm apart, which correlates to 11mm of coil activation. The overall distance between the 2 posts is 20mm. The eyelets at the ends of the springs are 3mm each which will consume 6mm of space. In addition, the diameter of the eyelets are 1.5mm (i.e., slop in the eyelet) which needs to be overcome before the spring are activated. This leads to 20mm – 6mm (eyelets) – 3mm (slop in the eyelet holes) = 11mm of activation. These springs were stored stretched in the laboratory in a salivary substitute material at 37°C and tested at time intervals of 4 weeks, 8 weeks, and 12 weeks on the Instron universal testing machine. The salivary substitute material used was Fusayama-Meyer artificial saliva. The composition of this solution, which closely resembles natural saliva is: KCl (0.4 gm/L), NaCl (0.4 gm/L), $CaCl_2{\cdot}2H_2O$ (0.906 gm/L), $NaH_2PO_4{\cdot}2H_2O$ (0.690 gm/L), $Na_2S{\cdot}9H_2O$ (0.005 gm/L), and Urea (1 gm/L)(20, 21). The solution was titrated to a pH of 6.5 using 5M NaOH.

3. Control springs (n=10): Ten of the pre-tested springs were analyzed on the Instron universal testing machine to identify the relationship between the two different machines used for pre- and post- testing.

Analysis

Initial and final pseudoelastic force-deflection curves were generated for each spring in the clinical, laboratory, and control groups. For all mechanical testing, each individual spring was stretched to approximately the same length for its post-testing on Instron as it was for its pre-testing on DMA. The maximum force attained by each spring at this length was used to compare the amount of force loss (force loss = maximum force reached by the spring at a given length pre-use – maximum force reached by the same spring at that same given length post-use).

At each time point, the loss in force was assessed using a one sample t-test to determine whether the mean loss was statistically significant. Unpaired t-tests were used to compare the control group force loss to the 4 week clinical data and separately to the 4 week lab data in order to evaluate for statistical significance beyond what is due to differences between different machines used for testing. One-way ANOVA was used to evaluate force loss differences between time points for the clinical springs, which had three mutually exclusive groups. Force loss differences between time points for the laboratory springs were evaluated using repeated measures ANOVA. Unpaired t-tests were used to compare force losses for the clinical groups at each time period to the laboratory values over that same time period. A linear regression model was performed to identify associations between the outcome variable of space closure/week and the predictor variables of sex, age, archwire type, slot size, initial coil stretch length, and individual coil stiffness (calculated by dividing the initial maximum force by the activation length of that particular spring).

RESULTS

Demographic

Space closure was carried out on 0.018 slot appliances (6 springs) and 0.022 slot appliances (24 springs) in addition to different wire dimensions: 18 SS archwire (7 springs), 18×25 SS (16 springs), and 16×22 SS archwire (7 springs). Overall, 11 patients (7 Female, 4 male) participated in this study (Average age 23.2 years, range 13–43 years).

Sample overall

None of the clinical springs showed signs of distortion upon retrieval so all 30 (10 per time point) were included in the analysis. During Instron testing at 4 weeks, two laboratory springs became distorted due to machine malfunction and were not used for analysis, giving a total of 13 laboratory springs for analysis. All 10 control springs were used for analysis.

Force

The initial and final pseudoelastic force-deflection curves for a single representative spring from the control, 4-weeks, 8-weeks, and 12-weeks groups are shown in Figure 3. The curve reveals a nearly flat plateau of constant force indicating a stress-induced martensite transformation during the loading. Interestingly, the DSC graph confirms that the GAC NiTi spring possesses a phase transformation in oral temperature (Figure 4).

The average and percent force loss values of all clinical, laboratory, and control springs are summarized in Table 1. All groups of springs (control, clinical, and laboratory) showed a statistically significant decrease in average force level from initial to final testing over each of the time periods evaluated ($p<0.01$). The control group springs showed only a 1.71% force loss. The clinical springs retrieved following 4 weeks of use showed an average force loss of 11.57% while those retrieved after 8 weeks of use showed an 18.88% force loss and after 12 weeks of use showed a 17.79% force loss. The laboratory springs showed an average force loss of 12.12% after 4 weeks of stretch, 17.36% after 8 weeks of stretch, and 19.44% after 12 weeks of stretch.

The amount of force loss experienced by the clinical and laboratory springs was significantly greater than that of the control springs, suggesting that the force loss was beyond an amount that can be attributed to differences between mechanical testing machines. Table 2 showed that the difference between the amount of force loss experienced by the control group springs and the amount of force loss experienced by the clinical and laboratory springs following 4 weeks of stretch was already statistically significant ($p<0.01$).

One-way ANOVA among the 3 mutually exclusive clinical groups revealed that there was a statistically significant relationship between the difference in force and the amount of time the springs were used ($p=0.04$). Table 3 shows pair-wise comparisons that indicated, on average, there was a statistically significant force loss between the 4 and 8 week time periods for the clinical springs of 21.90gms ($p=0.04$). However, the further force loss from the 8 to 12 week time points was not statistically significant ($p=0.93$).

Repeated measures ANOVA among the 3 time points for the single laboratory group revealed that there was a statistically significant relationship between the difference in force and the amount of time the springs were used ($p<0.01$). Table 3 revealed that, on average, there was a statistically significant force loss between the 4-week and 8 week time periods for the laboratory springs of 15.70gms ($p<0.01$). However, the further force loss from the 8 to 12 week time points was not statistically significant ($p=0.06$).

No statistically significant differences between the clinical groups at 4, 8, or 12 weeks and the laboratory group tested at 4, 8, and 12 weeks with respect to average force loss over the given time periods were demonstrated (Table 2).

Space closure

There were no statistically significant differences between the average amount of change in coil stretch length and the average amount of space closed for any of the clinical time periods ($p=0.47$ for 4 week group, $p=0.48$ for 8 week group, and $p=0.58$ for 12 week group). Figure 5 shows the average space closure distance for each time period (4 week group = 0.98mm, 8 week group = 1.70mm, and 12 week group = 2.71mm) with an overall average rate of 0.91mm/ month. The differences in space closure among three different time periods were statistically significant ($P=0.01$).

Table 4 summarizes linear regression data analyzing the association between space closure rate and the predictor variables of sex, age, archwire type, slot size, initial coil stretch length, and coil stiffness. None of these predictor variables demonstrated a statistically significant association with rate of space closure.

DISCUSSION

Although a limitation of this study was that initial and final testing of the springs was performed on two different mechanical testing machines, the unloading force-deflection curves for the control group springs on DMA and Instron were quite comparable (Figure 3A). The force-deflection curve at the fixed configuration such as the spring coil used here is the intrinsic property of the materials and should not be affected by the type of instrument used. However, the concern of the two machines is probably due to the different controls (load versus displacement for DMA and Instron, respectively) of testing, which could result in a tangential difference between two tests. In fact, the average difference in maximum force values between the two machines was only 1.71% (Table 1). Furthermore, there was a statistically significant difference between the force loss experienced by the control group springs (1.71%) and the force loss experienced by the clinical and laboratory springs (approximately 12%) following 4 weeks of stretch ($p<0.01$, Table 1). This suggests that the force decay experienced by the springs over a 4 week period was due to its intrinsic properties rather than measurement discrepancy between machines. Similar results are found for the 8 and 12 weeks clinical and laboratory springs. With the aforementioned validation, our outcomes could not introduce a false conclusion.

Our reported force decay following 4 weeks of use for clinical springs was 11.57% and for laboratory springs was 12.12%. This is in agreement with an *in vitro* study conducted by Angolkar et al. that showed an overall 8–20% drop over 28 days in force levels among various metal alloy coil springs.[13] A higher force decay percentage of 48% over a 22 week time period was reported in an *in vivo* study by Nightingale and Jones, which could be due to the fact that their values were measured with an intra-oral force gauge at the spring stretch lengths and thus likely included both the intrinsic force loss within the spring material itself as well as a large contribution from the fact that the coil spring length had decreased between time measurements due to space closure.[14] However, none of the springs tested in these studies were from the GAC company and since manufacturing conditions play a major role in force properties of these springs, the data is likely not very comparable with our study.

There were several *in vitro* studies by Maganzini et al, Manhartsberger and Seidenbusch, and Tripolt et al. that included data on the medium grade GAC Sentalloy springs used in our study.[5, 16, 22] However, the absolute value of our measured forces cannot be directly compared with their findings because all of our springs were not initially stretched to the full 12mm activation range that theirs were. It has been shown that the initial activation length of NiTi coil springs can have a significant influence on their force properties and effects both the length and absolute value of the constant force plateau regions of these springs.[23] However, an interesting finding of these studies that is in concordance with our results is that GAC NiTi springs do not exhibit constant force of the reported 150gm over the entire 12mm activation range as claimed by the manufacturer.[5, 16, 22] The super-elastic force plateau region occurs over a much more limited range than advertised.

Our study noted a significant drop in force of ~7% (clinical group) and ~5% (laboratory group) over the 4–8 week time interval with force levels appearing to maintain thereafter. Such findings agree with those of Nightingale and Jones who showed in a randomized clinical trial that NiTi coil springs experienced rapid loss in force over a 6 week period, after which force values leveled out. Nightingale and Jones hypothesized that this was because their initial activation lengths created force values higher than the super-elastic plateau and so it took a certain amount of space closure to reach a point where the springs were at a length to exhibit force within the constant plateau region.[14] The force-deflection curves in our study also demonstrated that our initial coil stretch lengths were beyond the constant force plateau region of these springs even though 23 of the 30 clinical springs we used were initially stretched less than the 12mm reported by GAC to be within the constant force plateau region and the average over-extension of the remaining seven springs was only 0.86mm. However, it is likely not the fact that the springs were initially activated beyond their force plateau that caused this initial rapid force loss in the springs over an 8 week time period because our initial and final measurements were taken with the springs at the same activation length and were thus unaffected by space closure. The initial force loss appears to be related to the intrinsic properties of the spring itself.

The fact that the clinical and laboratory data in our study showed no statistically significant differences with respect to force values is interesting since it has been noted that in most cases *in vitro* research gives a much different material profile from what is actually being used clinically.[18] Our findings agree with Wichelhaus, who demonstrated via *in vitro* thermo-cycling and mechanical micro-cycling tests, that the oral environment does not seem to be a major influence on the mechanical properties of NiTi coil springs.[23] Additionally, NiTi coil springs have been shown by Nattrass et al. to be unaffected *in vitro* by the specific environmental factors of water, Coke ®, or turmeric solution.[24] The biggest difference between the *in vivo* and *in vitro* environments that seems to concern researchers with respect to NiTi springs is the transient temperature changes experienced intra-orally because it has been shown that NiTi force properties are highly dependent upon temperature.[19] However, the force properties of the springs seem to be only altered while the spring is at that temperature rather than succumbing to some type of permanent deformation as a result of the transient temperature changes. Thus, temperature fluctuations only transitorily affect the force values supplied by the springs. Since our testing was performed at the constant temperature of 37°C for both clinical and laboratory springs, it makes sense they would demonstrate similar force values.

The rates of space closure obtained in our study, which averaged 0.91mm/month (0.23mm/week) were comparable to those reported in several other *in vivo* studies on NiTi coil springs, which ranged from 0.20–0.26mm/week.[14, 25, 26] The similar results in our study as compared to those of Dixon et al. was re-assuring as their method of space closure measurement with Vernier calipers on casts was likely much more accurate than our intra-oral Boley gauge measurements.[26] These studies were also better controlled than ours with respect to clinical and biomechanical variables. Since we still had similar reported space closure rates, it adds weight in support of our finding that none of the variables of sex, age, slot size, archwire, or initial coil stretch length are strongly associated with rate of space closure (Table 4). However, our study was really not powered to detect such relationships.

An interesting observation was the high degree of variability in spring constants supplied by each of the closed coil springs used in this study, especially since they were all obtained from the same manufacturer lot number. In order to demonstrate this, the maximum initial force value of ~300gm was applied to each spring and individual spring's activation length was recorded. The constant force of 300gm was divided by the individual activation length to obtain the spring constant. The spring constant ranged from 0.01717 to 0.03967 g/mm, and there was a great amount of variability between these values. It should be noted that each of the activation lengths is still within the 12mm activation range reported by the manufacturer to be on the constant 150 gm force plateau. This spring constant is quite a wide range of variability and is consistent with the findings of many other studies.[17, 27] There is a great deal of applicability of this information. Importantly, it indicates the need for tighter manufacturing guidelines to ensure advertised force levels are reached by the majority of springs. Further, clinicians need to be aware of the variability that exists between products because they are investing a system in what they believe to be delivering more biologically favorable forces than some of the cheaper space closure materials available such as elastomeric chain, which has been shown to close space at a comparable rate to NiTi coil springs.[14, 26, 28] It also highlights the fact that clinicians should really consider using an intra-oral force gauge when activating these springs to ensure that the force levels desired are actually being delivered.

CONCLUSIONS

- NiTi closed coil springs lose ~12% of their initial force following 4 wks of clinical use. An additional drop in force (~7%) occurs between 4 and 8 weeks of use, but force levels appear to stabilize thereafter. Therefore, force decays in a non-linear proportion to spring stretch duration.
- *In vivo* (clinical) and *in vitro* (laboratory) force loss data were not statistically different.

- Despite statistically significant decreases in force levels supplied by the NiTi closed coil springs, space closure appears to proceed at a rate of approximately 1mm/month.
- There is a significant diversity in force levels supplied by 150 gm GAC Sentalloy NiTi closed coil springs, even among springs with the same lot number.

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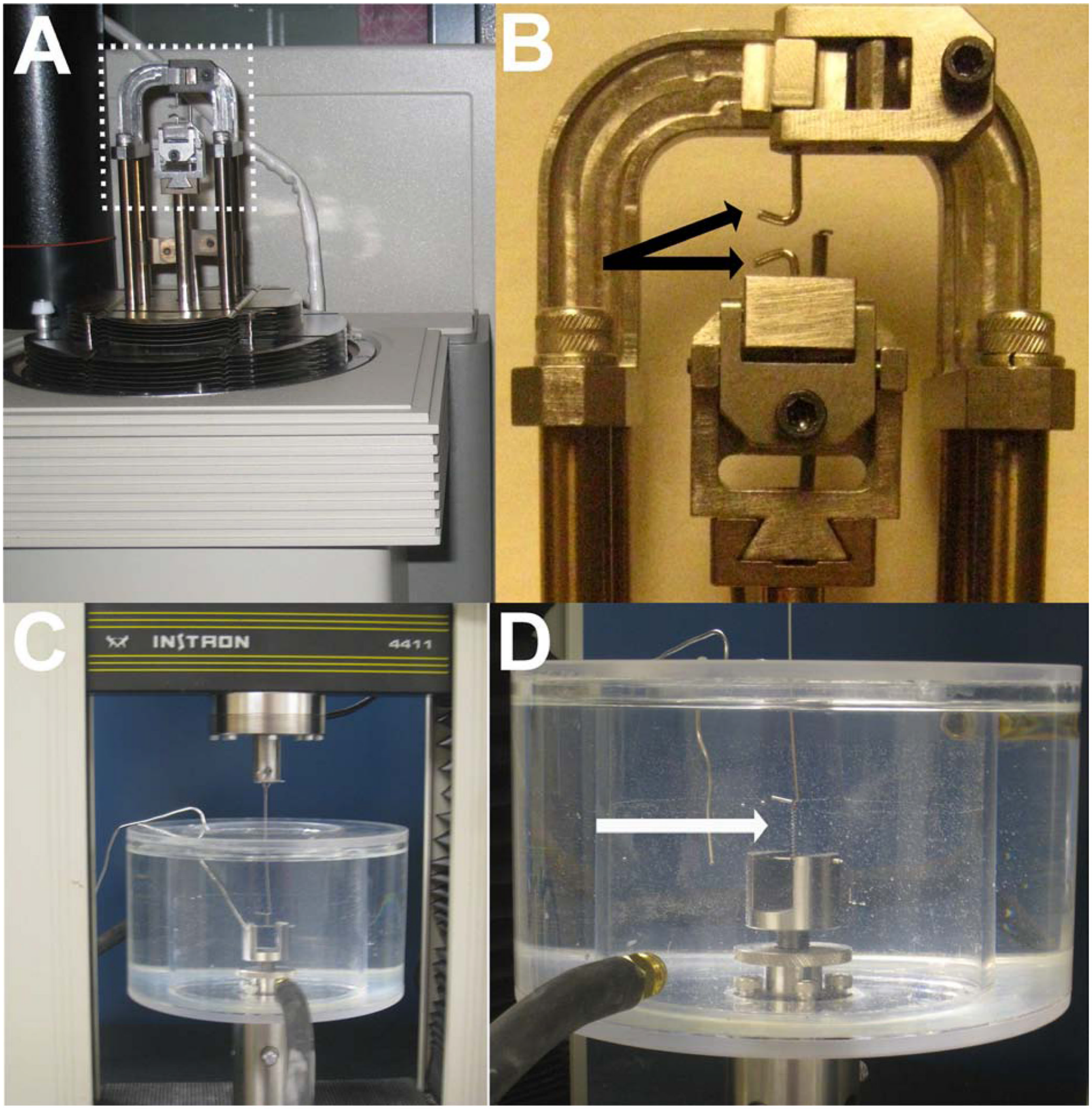

Figure 1.

A) Dynamic Mechanical Analysis (DMA) machine was used for initial mechanical testing for force level. B) A close-up (dash box in A) of the tension clamp with hooks used to attach coil springs for DMA testing. C) An Instron machine was used for post-spring use testing of force decay. D) The apparatus used for temperature controlled with the Instron.

Figure 2.
Example of coil spring attached during treatment of clinical group.

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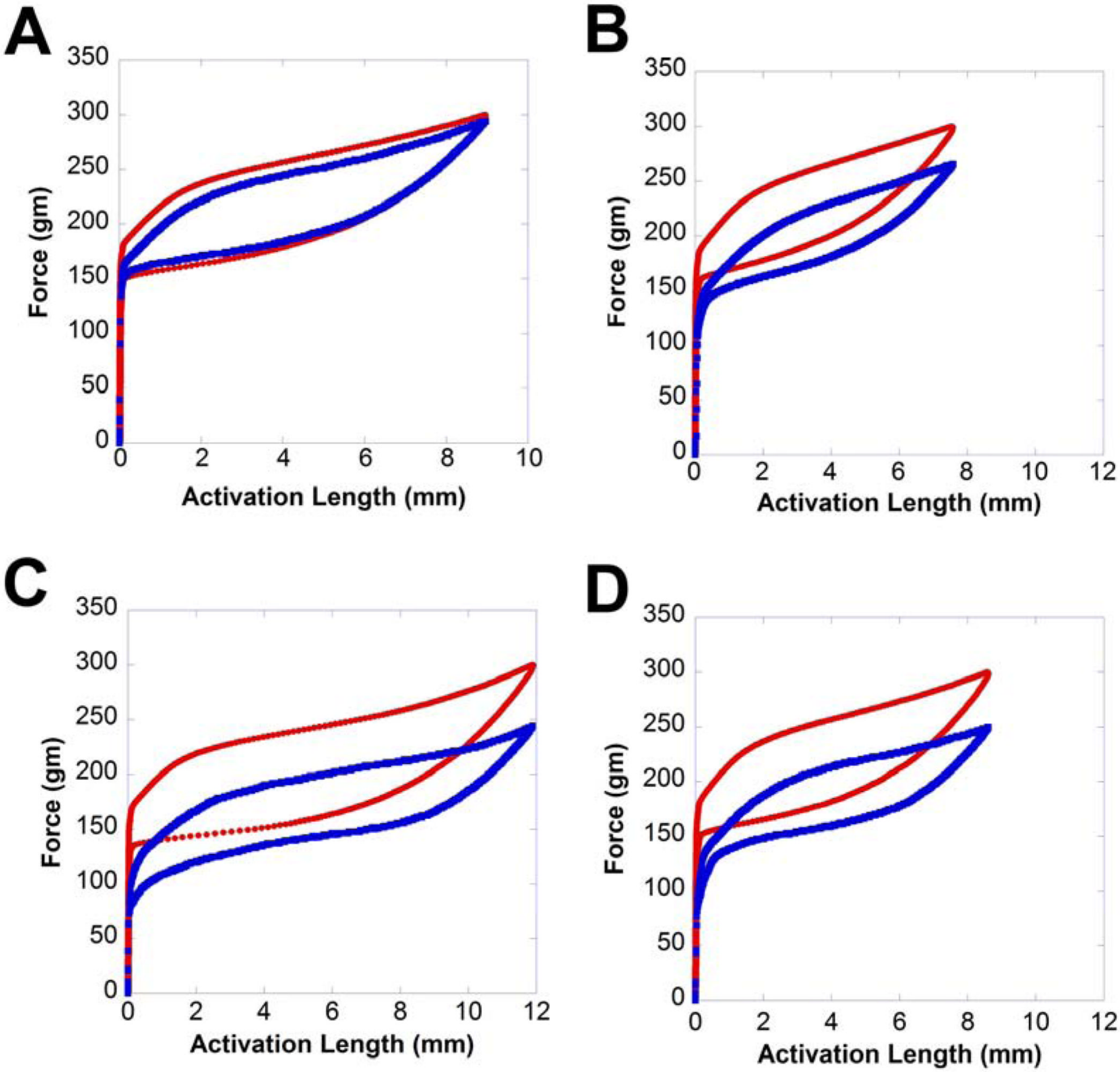

Figure 3.

Force-deflection curves plotting force levels versus activation length. Initial (Red) and final (Blue) force-deflection curves are shown for A) Control Springs, B) 4-weeks clinical use, C) 8-weeks clinical use, and D) 12-weeks clinical use.

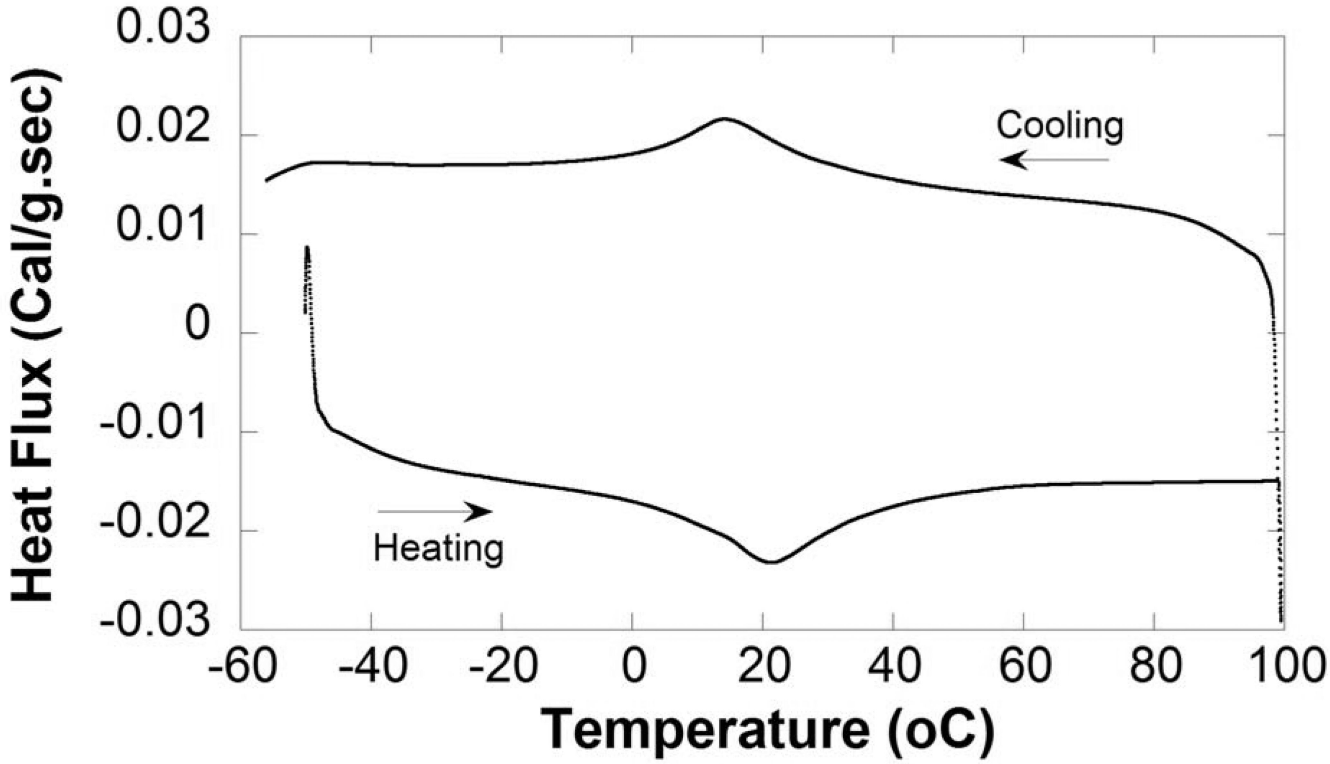

Figure 4.

Differential scanning calorimeter (DSC) analysis shows heat exchanges during both healing and cooling curves. The data indicates there is a phase transformation in the GAC NiTi spring used for this study.

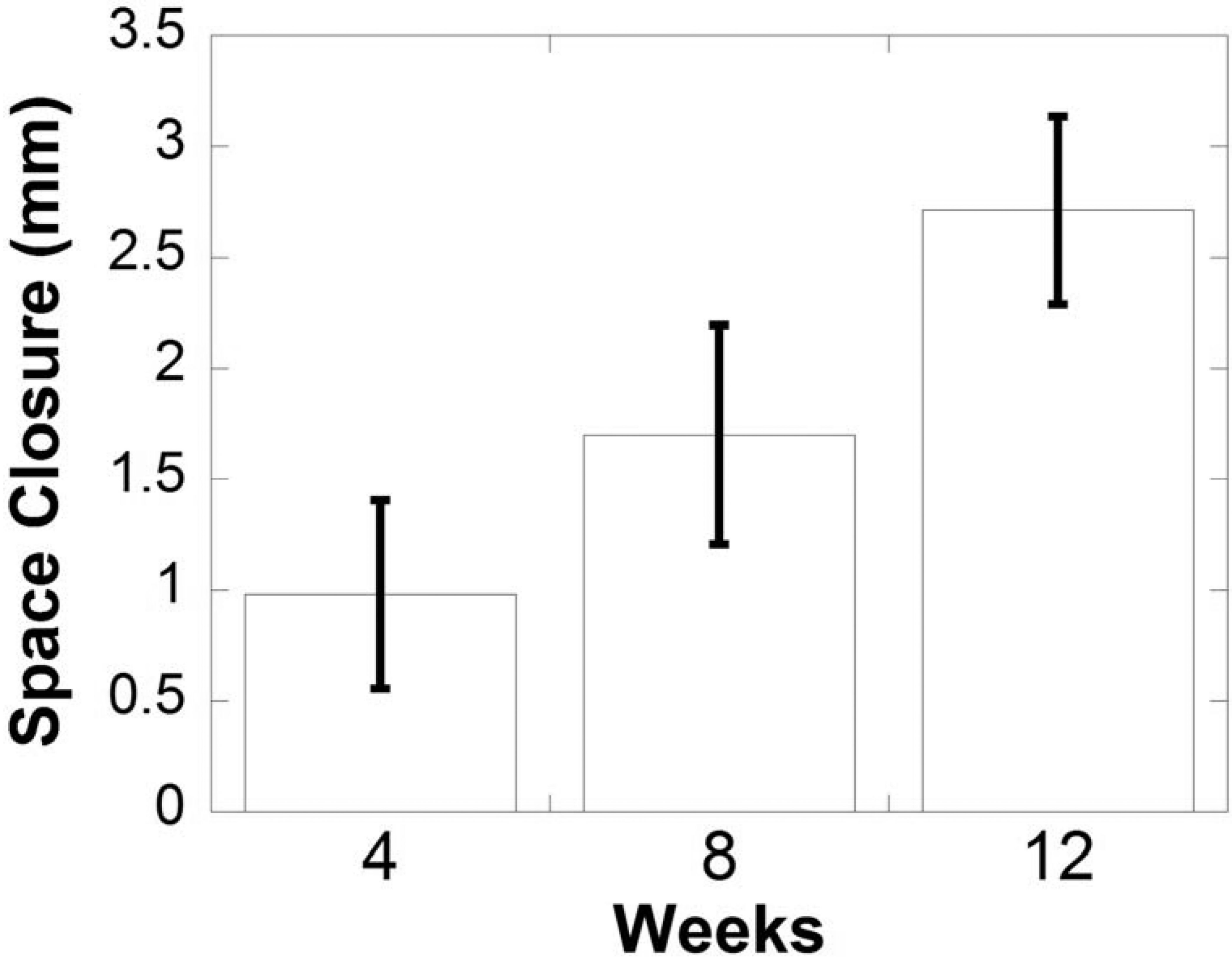

Figure 5.

Representation of average space closure amounts, with 95% confidence intervals, over each time period.

Table 1

Average and percent force loss for clinical, laboratory, and control springs over each time period.

	Time (weeks)	Mean Force Loss (gm)	S.D.	Percentage Force Loss (%)	P-Value
Control	0 (n=10)	5.12	3.53	1.71	<0.01
Clinical	4 (n=10)	34.69	19.99	11.57	<0.01
	8 (n=10)	56.59	18.93	18.88	<0.01
	12 (n=10)	53.33	19.10	17.79	<0.01
Laboratory	4 (n=13)	36.32	11.52	12.12	<0.01
	8 (n=13)	52.02	15.58	17.36	<0.01
	12 (n=13)	58.27	23.60	19.44	<0.01

Table 2

Mean difference in force loss between Control group, Clinical groups, and Laboratory groups at different time intervals.

Force Loss Comparison	Mean Force Difference (gm)	S.E.	P Value
Control vs Clinical (4 weeks)	29.57	6.42	<0.01
Control vs Laboratory (4 weeks)	31.21	3.79	<0.01
Loss at 4 weeks (Clinical Vs Lab)	1.64	6.61	0.81
Loss at 8 weeks (Clinical Vs Lab)	4.57	7.19	0.53
Loss at 12 weeks (Clinical Vs Lab)	4.94	9.16	0.60

Table 3

Force changes for clinical and laboratory springs between different time intervals. Standard Errors (SE) are reported for laboratory springs instead of confidence intervals.

	Time Period (Weeks)	Mean Force Difference (gm)	95% Confidence Interval		P Value
Clinical Springs (n=30)	4–8	−21.90	−43.35	−0.45	0.04
	4–12	−18.64	−40.10	2.81	0.10
	8–12	3.26	−18.19	24.71	0.93
Laboratory Springs (n=13)	4–8	15.70	1.56 (SE)		<0.01
	4–12	21.95	3.99 (SE)		<0.01
	8–12	6.25	2.99 (SE)		0.06

*Negative values for force indicate a loss over that time period while positive values indicate an increase in force.

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Table 4

Summary of linear regression data for the outcome variable amount of space closed(adjusted to mm/week) against the predictor variables sex, archwire type, slot size, age, initial coil stretch length, and coil stiffness (gm/mm).

Variable	F Value	P Value
Sex	1.06	0.31
Archwire Type	1.66	0.21
Slot Size	1.24	0.28
Age	0.01	0.94
Initial Coil Stretch Length	0.29	0.59
Coil Stiffness	1.10	0.31