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An In Vitro Investigation Into The Force Degradation Characteristics Of Nickel-Titanium Closed-Coil Springs In A Simulated Oral Environment With Simulated Tooth Movement.

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This investigation was undertaken to characterize the performance of various closed-coil nickel-titanium and stainless steel springs under conditions simulating clinical use. In toto, ninety samples of springs were extended on a testing jig which allowed the spring extension to be reduced at the rate of 2 mm per month, simulating clinical tooth movement. Force measurements were taken on the springs at ambient room temperature at specific intervals with a digital force gauge during the 12 weeks of the experiment. The apparatus was stored between force measurement sessions in 37 degree saline, and the springs were not removed from the apparatus during the experiment. In general, nickel-titanium springs were far more consistent than stainless steel springs in their force delivery during the course of the experiment, undergoing far less force degradation and delivering relatively constant, lower force levels. Stainless steel closed-coil springs delivered very high initial forces, which decreased rapidly and required re-activation after just 4 weeks, due to their short range of action. Lighter-force nickel-titanium springs were more consistent in force delivery than heavier-force ones, which in turn were far more consistent over time than stainless steel springs. It appears that nickel-titanium springs have the capacity to generate relatively constant force over a variety of ranges (depending on the initial force level of spring chosen) during simulated tooth movement in a simulated oral environment. This property may allow more physiologically suitable force levels to be applied during clinical tooth movement procedures.

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INTRODUCTION

There has been a lively discussion over the years in the orthodontic literature pertaining to the type and degree of force that should be applied to teeth undergoing orthodontic treatment.[27-48] A progression of investigators from Sandstedt[36] to Quinn and Yoshikawa[48] has attempted to clarify and quantify the events taking place in the peridontium and surrounding bone during tooth movement. It appears there is indeed an optimal range (albeit one with much inter-individual variability)[40] of force to induce such a movement; a minimum threshold necessary to begin the physiologic changes involved in tooth movement and a maximum beyond which tooth movement will slow or cease and perhaps result in tissue damage.[41, 48]

The ideal force system for tooth movement would seem to be one in which light continuous force within the so-called optimal range (70-140 g/cm^2 of root surface area) is provided to the tooth/teeth whose movement is desired while applying a sub-theshold force to teeth used as anchor units.[48] Many methods of applying force clinically have been utilized ranging from loops in the arch wires[1] and stiff stainless steel springs[4] to elastomeric materials[21-26] and more exotic alloys.[5-20] All of these, except for springs made of nickel-titanium alloy,[1,5,6,8,14] tend to develop initially very high force levels which subsequently decay rapidly in clinical use.[21-26] With the advent of nickel-titanium alloys, the possibility of using coil springs of this material to provide relatively constant force values over large deflections has appeared attractive.[1] Some authors have suggested this material may in fact be ideal for use in tooth movement procedures.[1,27,48] Previous studies have investigated the physical characteristics of nickel-titanium (NiTi) coil springs,[5,6,8,14] but none have examined their performance in a simulated oral environment using simulated tooth movement. This study was intended to provide data on the performance of such nickel - titanium springs in this environment.

A literature review of papers pertaining to NiTi coil springs, as well as other space-closure materials and techniques, is provided. The literature review will focus on materials currently used to deliver tooth-moving forces as well as related studies on the biology of tooth movement, friction, and root resorption as these are all inter-related subjects.

LITERATURE REVIEW

The introduction of nickel-titanium alloy coil springs into clinical practice invites comparison with other space-closure modalities, particularly with other alloys and elastomers. This review examines relevent papers. It is to be noted 'NiTi' is used thoughout this paper to refer specifically to 'super-elastic' nickel-titanium alloys, whereas 'Nitinol' and 'nickel-titanium' alone refer to the earlist, marstensitic-only alloy types.

Metallic Alloys

Pletcher[4] (1959) introduced the 'Pletcher' T-spring which is a 3/16 inch length of coiled stainless steel, of either .009" or .011" diameter wire with a .030" lumen. On one end 2.5 coils are reversed and the other end has a 2.75" straight wire extension. Pletcher recommended these coils, which are extended and ligated in place, be used for space closure when heavy continuous rectangular wires not larger than .021" X .025" are in place. In specific, he directs the loop in the coil be placed over the wire distal to the molar tube, and then the spring is pulled forward and bent around the hook on the archwire to activate it. It is re-activated as necessary. No mention is made of force levels.

Chaconas and Caputo[16] (1978) examined the force-extension characteristics of closed coil springs made of Hi-T, Permachrome and Elgiloy metals. They found increasing the wire diameter and decreasing the lumen diameter gave maximum force production for a given degree of activation. Springs with smaller wire diameter and larger lumen diameter remained longer in the ideal force range for tooth movement (about 250 g, reference Reitan[29]) and are therefore recommended for orthodontic use. Overall, they found incredible variation in the range and absolute value of the forces generated by various springs, and attempted to clarify the situation by providing a table to help clinicians better choose their springs.

Andreason and Morrow[15] (1978) commented on the characteristics of nickel-titanium wire with respect to an equivalent size of stainless steel. The shape memory, elasticity, and excellent working range of nickel-titanium wire were noted as was its superior energy storage capacity vis-a vis stainless steel. Since nickel-titanium wire has 1/2 the spring rate of stainless steel (spring rate = change in load/ change in deflection) for a given malocclusion, Nitinol produces lower, more constant and continuous force on the teeth than an equivalent sized stainless wire.

Burstone, Qin and Morton[17] (1985) characterized the then-new alloy, Chinese NiTi, noting it differed from conventional Nitinol in that it underwent little work-hardening, had an austenitic parent phase, and a lower martensitic to austenitic transition temperature than Nitinol. They compared Chinese NiTi with Nitinol and stainless steel, finding Chinese NiTi had springback that was 1.4X greater than Nitinol and 4.6X greater than stainless steel when measured in a bending test. In stiffness testing, they noted nickel-titanium alloys, unlike stainless steel and beta-titanium (TMA), don't have a linear relationship between bending moments and angular deflections. The stiffness of Chinese NiTi differs from conventional NiTi and stainless steel in that the unloading portion of its force/deflection curve drops rapidly from initial high force values, then enters a long range of near-constant deactivation force. Just before total deactivation, stiffness increases again. Also, the magnitude of force delivered increases for a given deflection if the wire is released and retied into a bracket. As well, Chinese NiTi ('super-elastic NiTi') is more resistant than Nitinol to time-dependent distortion. Overall, this wire was recommended for applications wherein one requires a wire type that delivers low stiffness and high deflection capacity.

Miura, Mogi, Ohura, and Hamanaka[12] (1986) examined the physical characteristics of Japanese NiTi. They noted unlike the original Nitinol wire manufactured by Unitek, Japanese NiTi exhibited characteristics of true 'super-elasticity' i.e. there is a large portion of the stress-strain diagram wherein the stress (force) remains fairly constant as strain (deflection) increases. This is due to the fact Nitinol is merely a work-hardened martensitic phase wire, whereas Japenese NiTi undergoes an actual austenitic to martensitic phase transformation upon deformation.

Buckthal and Kusy[13] (1988) investigated the effects of cold disinfection and sterilization on the characteristics of nickel-titanium alloys finding no corrosion and no change in physical properties.

Miura et al[14] (1988) examined the properties of Japanese NiTi alloy coil springs and compared them to commercially available stainless steel (Hi-T) and chrome-cobalt (Elgiloy) coil springs. An autograph machine with a load cell was used to stretch or compress the springs and perform force/deflection measurements. The results showed Japanese NiTi springs differed from the others in that their force/deflection diagram was flat, and further could not be predicted by any known formula (see Figure #1). Increasing the wire diameter and decreasing the lumen diameter had the effect of increasing force levels while decreasing the range of the superelastic activity. Martensitic transformation temperatures were varied with the finding that as it is elevated, the load value of the superelastic portion of the force/deflection curve is reduced. Clinical examples were given of the applicability of these springs to tooth movement, and it was noted it is possible to deliver nearly constant forces of supposedly ideal magnitude.

Mayhew and Kusy[11] (1988) subjected Nitinol and Titinal wires to dry heat, formaldehyde-alcohol vapour, and steam autoclave disinfection and sterilization, noting no change in mechanical properties or surface characteristics.

Boshart et al[7] (1990) tested the load-deflection characteristics of non-heat-treated and heat-treated chrome-cobalt-nickel alloy

(Blue Elgiloy) as compared to stainless steel (Hi T) in the form of open and closed coil springs. Ten samples of each were tested in air on an Instron machine. In general, the Elgiloy was stiffer than the stainless steel, especially if heat-treated. Stiffness increased as wire diameter and coil pitch angle increases and decreased as coil lumen diameter increased. A shorter spring was stiffer than a longer one.

Kapila, Haugen and Watanabe[10] (1992) examined the effects of dry heat sterilization and clinical reuse (sterilization plus reuse in a patient for 1 month) on the properties of martensitic (Nitinol) and austenitic (NiTi) nickel-titanium alloy. Dry heat sterilization did cause some change in the characteristics of the alloys, but it was not felt to be clinically significant. Clinical re-use did lead to an increase in stiffness in both NiTi and Nitinol wires and a reduction in superelasticity in NiTi wires.

Chen et al[9] (1992) bench tested in air 6 varieties of nickel-titanium and Chinese NiTi wire, finding the Chinese NiTi yielded much flatter stress/strain diagrams and exhibited superelasticity. The importance of the austenitic phase transformation temperature was noted in that a wire which does not undergo this transformation at mouth temperature will not be capable of exhibiting superelasticity or true shape memory.

Angolkar et al[8] (1992) looked at the force degradation of closed coil springs in vitro. Stainless steel, chrome-cobalt-nickel and three different types of NiTi alloy were tested using springs of two different lumen diameters and lengths. All springs were stretched to deliver an initial force of 150-160g , and force was recorded at intervals over a month. When not tested, the springs were stored on racks in salivary subtitute at 37 degrees. No simulation of tooth movement was performed. The results showed overall force decayed 8 to 20 % over a month and there was much variability between spring types. The NiTi springs, in general, did not perform any better than the other spring types. There was no attempt to control temperature of the springs at the time of force measurement.

Samuels et al[27] (1993) assessed the clinical rate of space closure in 17 subjects using an elastic module on 1 side and a Sentalloy (GAC) closed coil NiTi on the other over a period of 18 weeks. The module was replaced every 6 weeks, while the spring was left in situ. It was found the side with the spring showed a significantly higher rate of space closure than the other and no adverse effects such as tipping or tissue pile-up were noted when the spring was used.

Han and Quick[5] (1993) examined the properties of stainless steel and Japanese NiTi springs as well as elastomeric 'C-chain'. Fifteen 10 mm samples of each type were stored in 37 degree salivary substitute, statically stretched to twice their initial length. At 2 week intervals, they were stretched to 3 times their original length and returned to rest as force levels were recorded. The results showed stainless steel coil springs have an 'initial tension' i.e a certain force level must be exceeded to begin to open the coils; this seems to be imparted by the manufacturing process. NiTi coils and elastomerics don't seem to have this. The NiTi springs delivered the most constant force with the minimum amount of variance. It was noted, however, all materials tested did not give identical force readings in the elongation and relaxation phases e.g. a stainless steel spring stretched to 100% of its original length yields 640 g; the same spring elongated to 200% of its original length and relaxed to 100% yields only 200g of force. They speculated on the role of length cycling intra-orally and suggested an experiment be performed with simulated tooth movement.

von Fraunhofer et al[6] (1993) characterized the behaviour 6 open and 6 closed coil NiTi springs, as compared to similar stainless steel springs (Hi T). Force values were recorded by an Instron machine after the springs were elongated and allowed to quickly relax. Only deactivational forces were recorded. Results showed NiTi springs delivered approximately constant force over a 7 mm simulated rapid tooth movement, whereas stainless steel force levels degraded quickly. This experiment did not employ a simulated oral environment, nor were aging effects considered. Further, the rate of tooth movement was not representative of the usual clinical situation.

Elastomeric Materials

Andreason and Bishara[26] (1970) compared the performance of latex elastics and Alastik modules. A pilot study revealed they behaved similarly in water as in salivary substitute, and so their experiment was to age these materials in 37 degree distilled water. They found latex elastics lost 40% of their force on the first day, and then delivered relatively constant force for the next three weeks, whereas Alastiks had a force decay of about 75% on the first day, and were relatively stable for the ensuing three weeks. They recommended therefore the use of Alastiks which deliver an initial force some 4 times greater than that desired after the first day due to the 75% decay of force on the first day.

Hershey and Reynolds[24] (1975), referring to the work of Andreason and Bishara, noted the need to test elastomeric modules under conditions of simulated tooth movement, not just static conditions. They tested a total of 540 modules from three different manufacturers, 120 at a time, varying the initial interbracket distance in increments of 2 mm from 12 to 34 mm. All modules were aged in triple distilled water at 37 degrees Celsius, and the framework of the apparatus was closed at rates of both 0.25 and 0.5 mm per week. Two separate observers were used for measurements made with calibrated gauges and the experiment was run over a 6 week period. The results showed the elastomeric modules tested lost over 50% of their initial force value over the first 24 hours of the experiment. Thereafter, force decay continued at a reduced rate until, at 4 weeks, the force values had decayed to about 1/3 or 1/4 of their initial values. Simulated tooth movement, as expected, increased the rate of force loss, a tooth movement rate of 0.25 mm per week causing less rapid force decay than a rate of 0.5 mm per week. It was shown although initial force losses were high, the elastomers continued to generate force that was felt to be adequate to move teeth over a 4 to 6 week period.

Ash and Nikolai[23] (1978) compared the force degradation characteristics of elastomeric chain in a water bath to that which occurs intra-orally. They found the intra-oral elastomers underwent a greater and more rapid force degradation than those in the water bath.

Bertl and Droschl[21] (1986) submerged typical intra-oral elastics in 37 degree saline solution and showed a significant reduction in force over the first 3 hours, and then no real change up to 8 hours.

Kuster, Ingervall and Burgen[22] (1986) performed in-vitro and in-vivo force assessments of elastomeric chain up to 4 weeks finding after an initial stretch of 2 times their initial length, the greatest force loss in vitro was 10 - 16 % during the first 2 hours, increasing up to 25-30% at 4 weeks. The in vivo results showed an even greater decline in force, up to 50% at 4 weeks. None of the elastomers had force levels which declined below 100g at 4 weeks.

Lu, Wang et al[25] (1993) studied the force degradation characteristics of elastomeric chain stored in a simulated oral environment and undergoing simulated tooth movement. Various elastomeric chains were stretched and stored in 37 degree water at pH 7 for 6 weeks; length of stretch was decreased by 0.5 mm per week to simulate tooth movement. In general, their results mirrored those of previous investigators in that there was a rapid initial force decay in the first hour and about half of the initial force level was lost at 4 weeks. The greater the initial force, the more the force decayed.

Force Levels and Clinical/Histologic Effects

Previous investigators have demonstrated there perhaps are optimal levels of force magnitude and duration with respect to orthodontic tooth movement.

Sandstedt[36] (1904) assessed histologically the effects of orthodontic force on dog teeth, noting the use of excessive force led to what he termed "undermining resorption" of bone adjacent to the teeth, rather than continuous frontal resorption on the compression side of the alveolus. He also noted the formation of bone spicules on tension side of the alveolus.

Oppenheim[37] (1911) replicated these experiments, but claimed to find overall deposition of bone all around the tooth, which was then followed by selective resorption on the compression side of the tooth. In a 1944 paper[38], he relates his investigation of the histologic changes around monkey incisors which were tipped labially via three methods i.e. coil springs of 360 g and 240 g force, as well as a stopped arch wire to increase arch length. Histologic sections were taken during active tooth movement, as well as after wires were removed and teeth allowed to rebound. He found osteoclasts were mobilized to start resorbing the bone on the compression side of the alveolus shortly after the application of force to the tooth and these cells continued to work for 4 days once mobilized. Further, osteoblasts at the same time were laying down osteoid on the tension side of the alveolus. If excessive force is used, the blood vessels and collagen fibres on the tension side are torn, and no osteoid is formed there. This was felt to be a contributing factor to relapse, as he noted osteoid seems relatively resistant to immediate resorption and so helps to hold the tooth in its new position. As well, the periosteum on the compressed side is crushed and strangled resulting in tissue necrosis and the disappearance of cellular elements and undermining not uniform frontal resorption is the consequence. Noting the forces used (240 or 360g) were too strong, and the development of root resorption areas was dependent also on duration of force, Oppenheim recommended the use of light, intermittent forces. This would allow rest periods during which the body could repair damaged areas of root and bone. Stating all forces caused some tooth damage, he went on to say force measurement clinically was a waste of time since there was such variation in individual response to force.

Schwarz[35] (1932) criticized Oppenheim's 1911 findings attributing them to an improper assessment of histological preparations. After performing his own experiments in which he observed histologic changes around three dog premolars which were tipped buccally by a calibrated spring, he concurred with Sandstedt and devised the concept of 4 degrees of biologic effects due to orthodontic force:

1. <u>1st degree</u>: a force of short duration and small magnitude will result in no appreciable biologic reaction.

2. <u>2nd degree</u>: a force of a certain minimum duration which produces pressure which is less than that of the periodontal blood capillaries will cause rapid bone resorption at the pressure site. There will be a repair of the periodontium when the pressure stops.

3. <u>3rd degree</u>: the applied force generates a pressure within the periodontium which is greater than the blood capillary pressure, and thus cell death and 'hyalinization' of the periodontal membrane occur, as well as resorption of the root. The periodontium will restore itself when pressure is released, but there may be permanent root damage.

4. <u>4th degree</u>: an excessivly strong force crushes the periodontium and causes extensive undermining bone resorption and root resorption. Even when the force is removed, there may be ankylosis of the tooth or pulp death.

Schwarz recommended therefore that the pressure generated by applied orthodontic forces should not exceed 20 - 26 g/cm^2, which he felt to be mean capillary blood pressure.

Reitan[30] (1947) examined histologically the effects of tipping versus bodily tooth movement on dog teeth finding tipping induces more root resorption than bodily movement, presumably because the pressure gradient along the root surface was more even in bodily movement. This held true as long as light forces are used (45-55 g). That is to say, for a given force level, " continuous bodily movement of teeth seems to imply root resorptions to a lesser degree than in cases where teeth are moved with approximately the same or even lighter forces, but not bodily." He also noted as teeth moved labially "the apposition at the outer labial bone surface was of a thickness decreasing apically and approximately proportional with the degree of resorption from inside of the alveolar bone wall."

Reitan[29] (1951) reviewed all available histologic literature as well as conducting his own histologic investigations of various tooth movements on human and dog subjects. He disagreed with Oppenheim in that he saw only osteoid deposition and an increase in cell number on the tension side and bone resorption and a decrease in cell number on the pressure side only. He agreed with Opppenheim that this osteoid seemed more resistant to resorption than nearby alveolar bone, and that if resorbed it occured from the rear. Since interrupted forces with a recovery period allowed the formation of osteoid on the previously compressed side, this would result in slower tooth movement; therefore he favoured continuous forces.

Storey and Smith[32,33] (1952) introduced the concept of differential force thresholds for the optimal movement of various teeth, stating that " it may reasonably be expected that there is an optimum force (or range of forces) which should be used to bring about this change in position in a reasonable time, with a minimum of damage to tissues and with a minimum of inconvenience to the patient." They noted the same force applied to two different teeth with different root surface areas will cause different pressures to be produced in the periodontium of the teeth. For example, a force which generates the supposed optimal pressure of 20-26 g/cm^2 on a cuspid will produce a lesser pressure in the periodontium of a molar because the force is distributed over a larger surface area. Thus the cuspid will move, but the molar will not because the force in its periodontium is below that required to induce bony remodelling. Conversely, an excessively large force will exceed capillary pressure in the periodontium of the cuspid and induce cell death and hyalinization, followed by slow undermining as opposed to frontal bone resorption. The same large force will induce pressures which are in the optimal range for movement of the molar, and so the supposed 'anchor' tooth will move more rapidly than the cuspid that one intends to retract! They conducted clinical experiments in which they compared the tooth movements which occured when light (175 -300 g) and heavy (400 - 600 g) helical springs were used to retract cuspids via a sliding yoke arrangement. In testing their springs, they noted a huge range of force values generated along with a rapid decay in force applied to the teeth. They commented that the existence of 'individual variation' had in the past been used as an excuse for failure to standardize force levels in orthodontic appliances. The results showed the light springs consistantly gave a rate of space closure of about 1 mm per week due to distal movement of the cuspid. The heavy springs caused essentially no movement of the cuspids but did cause mesial movement of the molars until the force generated by the spring had declined to a level of about 250 g, at which point the molars stopped moving and the cuspid began to move. This behaviour seemed to hold for all of the patients tested. Thus, they recommended that force levels in the range of 150-200 g are optimal for cuspid retraction, especially if the force is continuous. They felt that the 150 g minimal force threshold indicated the cuspids could withstand a certain level of pressure before bone resorption began. This work stressed the point the teeth are not tent pegs and we must be aware of the force levels we apply and the biologic response to them.

Begg[47] (1956) said light wires applying light forces produce the " least tooth mobility, least pain, and fastest tooth movement." If heavy forces are used, it was felt to be possible to apply the Storey/Smith concept of differential force for anchorage i.e. anterior teeth could be used in theory as anchors to protract posterior teeth. By using light elastic force of 150 - 200 g, he claimed to be able to retract cuspids and incisor teeth into extraction sites without any molar movement at all.

Reitan[28] (1957) said there are three main factors to be considered with regard to orthodontic force, all of which are linked:

1. individual variation in tissue response
2. the type of force applied
3. the mechanical principles involved

With respect to individual variation, Reitan notes there are many histologic variations between individuals of the same age group as well as between age groups, such as a decrease in periodontal cellularity and increase in bone density in older individuals. Noting the lag time of one week before cellular proliferation and osteoid formation on the tension side of a tooth, he stresses that high initial forces are not productive, Further, he states until the lamina dura on the compression side starts resorbing, the application of high forces will hasten the formation of necrotic cell-free areas here and thus slow movement. He recommends light initial forces for tipping movements, about 25 g for adults and 40 g for young patients, which can be increased later.

In discussing the type of force as applied to tipping of teeth, he distinguishes between continuous, interrupted continuous and intermittent forces, stating intermittent forces (i.e. a spring which delivers a force which decays to zero) are best, as they allow time for cellular elements to infiltrate the site of compression. Functional intermittent forces of 70 - 100 g may cause cell-free areas to form, but these are less extensive and shorter in duration than in continuous tooth movement. Thus removable functional appliances work best for tipping teeth. He notes identical force levels will produce more hyalinization in tipping than in bodily movement, due to the different surface areas of periodontium available to dissipate the force. He reiterated the Storey/Smith concept of differential force effects on teeth with different root surface areas and recommended for continuous bodily tooth movement force values of 150-250 g to retract upper cuspids, 100-200 g to retract lower cuspids, and 25 g to extrude incisors.

Reitan[31] (1960) reviews histologic investigations into tipping, bodily, and rotatory tooth movements. He noted an increased likelihood of root resorption in tipping versus bodily movement. He noted a difference in bone resorption/apposition patterns, in that tipping caused force concentration at the alveolar crest and root apex inducing bone resorption in the areas of compression and deposition in the areas under tension while bodily movement induces more even resorption all along the bone surface on the compression side with deposition occurring evenly on the tension side. With regard to tooth tipping, he noted there was a residual tension in the periodontal tissues subsequent to such movements and enough pressure could be generated after removal of tooth-tipping force to induce resorption in the area previously under tension, thus contributing to relapse. he stated that " it was felt, following rotation, tension and displacement of supra-alveolar structures may persist even after retention. Early treatment or over-rotation may, to a large extent, prevent relapse tendencies."

Jarabek[43] (1960) felt a force of 28-110 g applied to a tooth would produce equal cellular activity on the tension and compression sides of the root. This was felt to yield optimal tooth movement.

Stoner[59] (1960) felt, although it was difficult to quantify the forces which are applied to individual teeth, an optimum force for cuspid movement was 60-180 g.

Burstone[42] (1965) recommended a force of 150 g be applied to tip cuspids distally via his 'cuspid retraction assembly'.

Weinstein[44] (1967) used onlays on premolars to increase the resting pressure of the buccal musculature on these teeth and found significant tooth movement with increases in applied forces as low as 1.68 g, thus demonstrating that low level, continuous forces can move teeth. The duration of applied force seemed to be the critical factor, not the intensity.

Tacy[45] (1968) measured the rate of tooth movement while applying various forces to retract cuspids via closing loops on a straight arch wire. He disagreed with Storey and Smith, finding greater force resulted in more rapid space closure and there was essentially a straight-line relationship between force and rapidity of space closure over a range as wide as 50-1500 g, with space closure occuring more rapidly in the maxilla than in the mandible. He did not find forces greater than 150-200 g caused decreased cuspid movement and increased molar anchor movement. He did find cuspids were retracted with forces less than 150 g, again disagreeing with Storey and Smith. No comment was made on any potential relationship between force levels and root resorption.

Hixon et al[39] (1969) examined the Storey/Smith, Begg and Reitan concepts of an 'optimal' tooth movement force using clinical and radiographic data to assess the relation between cuspid and molar movement, and force. They attempted to remove the effects of cuspid rotation and tipping in their appliance design and found higher forces per unit root area, up to a level of 3 or 4 g/mm^2, increased the rate of biologic response. They noted tip-back bends merely compensated for crown tipping induced by wire flex or improper bracket position. They felt there was more rapid crown movement with light wire appliances because tipping caused a large increase in pressure at the alveolar crest. In summary, they felt for total forces of 300 g or less, the average rate of tooth movement increased as the load per unit root area increased whether the tooth is tipped or moved bodily. There was no accounting of the potential role of friction done in this study, nor was root resorption addressed.

Hixon et al[40] (1970) followed up their 1969 paper with a clinical study employing better control of tipping and rotation. They found two distinct phases of tooth movement:

1. initial mechanical displacement, and
2. a delayed tissue response

It was felt that individual variation with respect to root surface areas, response times, etc. were all much more important than variations in force levels used to move teeth.

Paulson et al[46] (1970) used laminography to evaluate cuspid retraction and molar anchorage loss, finding if a force of 50-75 g is used to retract a cuspid (using a .018:" bracket and .016" stainless steel wire) , there was no molar anchorage lost with average cuspid retraction being 3.9 mm. They commented on the importance of appliance friction and patient compliance as variables.

Buck and Church[34] (1972) applied tipping forces of approximately 75 g to premolars in human subjects, and extracted them at 7, 14, 21, and 28 days. These were then examined histologically. After 7 days of tipping force application, they noted ischemia and cell death in the compressed areas of periodontal ligament, with the formation of a 'cell-free' area and undermining not frontal resorption. Any tooth movement occuring was due to periodontal ligament compression or bone bending. At 14 days, a breakthrough into bone marrow spaces gives a rapid restoration of cellular elements, resulting in frontal resorption along with osteoblastic activity and resultant immature bone formation. Minimal compression of the periodontal ligament was noted, and patent blood vessels were seen. At 21 days, reorganization of the periodontal ligament and alveolar wall were noted, with significant osteoblastic and fibroblastic activity and minimal osteoclasis. The appearance of lateral root resorption was noted. The 28 day specimens showed almost complete reorganization of the periodontal structures. They suggested that with this force level there was a minimal lag period, noting undermining resorption as early as 7 days after force application, and stating “frontal resorption and tooth movement through bone should be a clinical possibility after 7 days.” Periodontal ligament cell loss and collagen changes due to compression-induced ischemia were felt to be reversible phenomena.

Boester and Johnston[41] (1974) examined the effects of force levels on perceived pain and rate of tooth movement by applying clinically 4 different forces (60, 150, 240, and 330 g), one in each quadrant of a 4 bicuspid extraction case, to retract cuspids. Friction played no role as independant cuspid retraction springs were used. They found space closure proceeded equally rapidly with force levels of 150, 240, or 330 g, whereas a 60 g force produced slower movement i.e there is a range of optimal bone resorption and force may play a role only at the lower levels. They found no support for the concept of differential force in that anchorage loss was independant of the forces used. There was no difference in pain levels reported at the various force levels. They criticized the experimental design of Storey and Smith noting by using a continuous wire from the molar to the incisors, bypassing the cuspids, they allowed binding of the molar anchors on the wire due to wire flex and so it may have been impossible for them to get mesial molar movement except at high force values.

Quinn and Yoshikawa[48] (1985) reviewed the literature relating to tooth movement rates and applied force and found overall tooth movement behaviour is best represented by a linear relationship between the magnitude of stress and the rate of movement. This plateaus after a certain level with the result further increases in force do not cause an increase in rate of tooth movement. They surmised this was like many other bodily responses in that it was a 'saturation effect' i.e. the body can only mobilize so many osteoblasts and osteoclasts at a given time and so excess force is unwarranted.

They felt that the best estimate of maximally efficient canine retraction force from clinical data is 100-200 g, equating to a compressive stress on the cuspid root of 70-140 g/cm^2. They recommend to minimize anchorage loss, an appliance should be used which delivers constant forces in this range.

Proffit[1] (1993) states the ideal force to slide a canine tooth distally is about 200 g, of which about 1/2 is used to overcome friction. A NiTi coil spring is preferred over stainless steel springs or elastomeric materials because these can offer a constant versus a rapidly decaying force.

Graber[2] (1994) states "from a clinical point of view, an optimal force is one that produces a rapid rate of tooth movement without discomfort to the patient or ensuing tissue damage (alveolar bone loss and root resorption, in particular). From a histologic point of view, an optimal force is one that (1) basically maintains the vitality of the tissue throughout its length and that (2) initiates a maximum cellular response (apposition and resorption). Optimal forces therefore produce direct resorption of the alveolar process. Since optimal forces require no period of time for repair, it appears that such forces can be made to act continuously." Further, with respect to patient discomfort, he mentions clinical studies have demonstrated that "not only is a higher degree of pain evident with a heavy force, but the total number of days in which an abnormal pain response can be elicited is increased."

Friction

Stoner[59] (1960) said "recognition must always be given the fact that, because of appliance inefficiency, sometimes applied force is dissipated by friction or improper application, and it is difficult both to control and to determine the amount of force that is being received by the individual tooth."

Riley et al[62] (1979) found steel ligation of archwires generated more frictional resistance than did elastomeric ligation, especially in water where the steel ligatures corroded.

Frank and Nikolai[49] (1980) investigated the relationship between orthodontic brackets and archwires. They found at small wire/bracket angulations bracket width and ligation force were the dominant influences on friction. As this angulation increased and binding occured between the wire and bracket, the angulation of wire to bracket became the dominant factor with regard to friction. At very high angles, wire shape and stiffness exerted more influence on friction.

Peterson et al[60] (1982) looked at the influence of bracket interwing distance, wire/bracket angulation, and wire type on friction. At small wire/bracket angulations, there was essentially no difference between stainless steel and Nitinol wires, but as angulation increased, stainless steel showed a greater increase in friction than Nitinol. They stated large rectangular Nitinol wire could be used during space closure without an increase in frictional resistance. Bracket width had no effect on friction.

Garner et al[58] (1986) looked at frictional forces during simulated canine retraction on nickel-titanium, TMA, and stainless steel wires, finding stainless steel had the least friction, TMA the most.

Stannard et al[50] (1986) evaluated variation in friction under wet or dry conditions concluding artificial saliva actually increases friction between brackets and various wires due to the fact polar liquids like water increase adhesion between polar materials thus increasing attraction between the materials. In the wet state, TMA or stainless steel on stainless steel brackets yielded the least friction.

Drescher et al[61] (1989) conducted a thorough assessment of the frictional forces between a bracket and archwire, noting there are four distinct phases in the guidance of a tooth along an archwire:

1. Before the application of a mesio-distally directed force and at the completion of the levelling stage, the archwire is passive in the slot.

2. As force is applied, the tooth begins to tip and translate.

3. Continuous force application causes elastic deformity in the archwire; the load at the contact point increases, as does friction. Elastic deformation in the wire induces antitip and antirotation movements of the tooth.

4. If the forces are unbalanced, permanent deformation of the wire can occur.

Stating the force needed to move a tooth is equal to the sum of frictional force and biologic retarding force, they feel that any applied orthodontic force must be at least two times that needed just for the biologic response, in order to compensate for friction. They also state the factors most affecting friction in decreasing order are:

1. biologic resistance
2. wire surface roughness
3. bracket width (narrower causes more friction)
4. elastic properties of the wire (friction increases slightly as elasticity increases)

Angolkar et al[51] (1990) studied friction between ceramic and stainless steel brackets and 4 wire types (TMA, NiTi, CoCr, and stainless steel). Friction increased as wire size increased with rectangular wires causing more friction than round wires. TMA and NiTi caused greater friction than stainless steel or CoCr. In general, friction increased as wire diameter increased, and narrow single - wing brackets caused less friction than double - wing brackets. Using .022" slot siamese brackets, they found there was 222 g of friction to overcome.

Kusy and Whitley[55] (1990) evaluated the effects of surface roughness on friction, determining it is perhaps not the best indicator of friction. They felt one must consider the reactivity of the materials involved, as well as relative hardness and softness.

Pratten et al[56] (1990) examined the frictional characteristics of stainless steel and nickel-titanium rectangular wires in both stainless steel and ceramic brackets, finding stainless steel wire caused less friction than nickel-titanium wire, and ceramic brackets caused higher friction than stainless steel ones. The worst combination was nickel-titanium wire in ceramic brackets. They also found artificial saliva increased friction.

Tanne et al[57] (1991) used an experimental jig to simulate cuspid retraction and measured tooth movement and microscopic surface changes in archwires and brackets, both stainless steel and ceramic. They found they achieved greater tooth movement with stainless steel brackets and ceramic brackets scratched the wire more than did stainless steel.

It is to be noted that all of the cited friction experiments investigated essentially static, planar friction; few if any assessed the effects on friction of the 'jiggling' of teeth which inevitably occurs in the clinical situation.

Root Resorption

Apart from potentially causing movement of abutment teeth and increased patient discomfort, Schwarz[35] (1932), Oppenheim[38] (1944), Reitan[28] (1957), Wainwright[63] (1973), and Remington[64] (1989) all suspect the use of high force levels may be related to an increase in root resorption, particularly in those cases where the root contacts the cortical bone plate.

Henry and Weinmann[65] (1951) stated in most instances cemental root resorption appears to cease and repair takes place when the traumatic stimulus (e.g. excessive force) is removed.

Reitan[31] (1960) noted movements which distributed force over a larger surface area rather than a small area tended to produce less root resorption.

Proffit[1] (1993) noted continuous forces of high magnitude prevent repair processes and cause rapid metabolite build-up and cell death. This is felt to result in increased tooth mobility, pain, and root resorption.

Graber[2] (1994) also concurs force magnitude may play a critical role in root resorption.

Brezniak and Wasserstein[66] (1993) provide an excellent review of the many factors which can affect the occurence of root resorption, including high force levels.

METHODS AND MATERIALS

Two testing jigs were fabricated in a fashion similar to that of Lu et al[25] to allow for the testing of force decay over time of a variety of stainless steel and nickel-titanium springs (see Table #1) with simulated tooth movement. Two pieces of plexiglass, into each of which had been inserted 46 stainless steel pins (23/side), were held at a fixed distance apart from each other by two threaded galvanized rods. The degree of separation could be varied by adjusting the lock nut/washer/butterfly nut assembly supporting the upper portion of the apparatus (see photographs #1, #2 and #3).

Ten different types of space-closing springs (two stainless steel and eight nickel-titanium) representing five major orthodontic supply companies were purchased from available commercial stock (see Table #1). Nine samples of each type were initially mounted to provide an initial elongation of twice their resting length (for the nickel - titanium ones) or three mm (for the stainless steel Pletcher springs) as per manufacturers suggestions. Special stainless steel wire hangers (.030" stainless steel) of uniform length were made to suspend all of the springs from the top member of the apparatus while other stainless steel connectors of lengths specific to each particular spring length and type were made to render the desired initial spring elongation. Initial separation between the two plexiglass blocks was 100 mm and this was closed down at the rate of 0.5 mm each week after force readings were taken, to simulate clinical tooth movement.[24,41] The final readings at the end of the twelve week period were taken at 94.5 mm of separation between the plexiglass blocks.

The two jigs were immersed in 37 degree distilled water for the duration of the experiment (photographs #4 and #5).[26] When readings of force values were to be obtained, the jigs were removed from the water and allowed to equilibrate thermally with ambient room temperature which was recorded at each measurement session (mean temperature 21 degrees Celsius).

All force measurements were made using a hand-held digital force gauge (Ametek Acu Force Cadet, Mansfield and Green Division, Largo, Fla. U.S.A.) with a full scale deflection of 1000 g, rendered in 1 g increments (photograph #6). Force values were recorded at the point at which a vertical force, exerted by the force gauge on the hook of the wire suspending the spring, caused the wire hanger to break contact with the stainless steel pin on which it was suspended. This was done using visual inspection of the area involved. All measurements were repeated at least twice and if there was a variance of greater than 10 g between measures on one spring, a third measure was taken. These were subsequently averaged. Initial force values were recorded at the time of spring placement and subsequent measures were made at intervals of 1 hour, 8 hours, 24 hours, 3 days, and then weekly until 12 weeks had passed[8]. Due to their short range of activation (3 mm), the stainless steel springs were reactivated after 4 weeks by having their lower hanger wires replaced by new ones which were 3 mm shorter than the first. The nickel - titanium springs were never removed from the apparatus or reactivated. Both types of springs were so handled to replicate clinical use.

RESULTS

Raw data and various statistical manipulations are included as an Appendix . A 2 X 17 randomized block design Anova was used to look at Springs (F=339.78; df=9,80), Time Periods (F=170.84; df=16,1280) and Springs by Time Period interaction (F=54.65; df=144,1280). All three were significant at P<.0001 level of significance.

The significant effects for Springs indicate the mean pressure for each type of spring differed. Table #2 summarizes the results of comparing all of the means, two at a time, using Scheffe tests.

The significant effect for Time Periods indicates that the pressure (force) changed over time and the significant interaction tells us the force changes for the various springs were not parallel i.e. some springs changed the force levels that they delivered faster than others. Figure #1 demonstrates this graphically by showing pressure (force) changes over time.

One of the principal goals for this study was the assessment of the capacity of various spring types to deliver relatively constant forces in what is considered to be a physiologically appropriate range for tooth movement. To assess this, an analysis of simple effects was done on each spring. This essentially was a one - way ANOVA which looks at each spring separately, but uses the error term for the complete ANOVA as well as its degrees of freedom. The results are presented in Table #3 starting with the least significant F value (GAC 100 g, F=0.23) and increasing to the largest F (Unitek 12X30 SS, F=511.4). That is to say, only two springs did not show a significant force degradation over the experimental time period, GAC 100 g and Masel 100 g (both nickel - titanium springs).

The mean force exerted by the GAC 100 g spring over twelve weeks was 113.7 g and by the Masel 100 g it was 79.5 g. The RMO nickel - titanium spring was the third most reliable, yielding an average force of 183.2 g. It did, however, seem to degrade significantly (statistically) over the experimental time period. Figures #2, #3, and #4 utilize an expanded ordinate scale to demonstrate variations in force levels over time which are not readily observable in Figure #1 with its 1000 g ordinate range. This demonstrates that while the change in force in the RMO springs is statistically significant, it is still less than 50 g.

The Unitek stainless steel springs were included in the study to provide a comparison with conventional stainless steel spring space - closure modalities, at least in terms of force degradation over time. It is apparent these springs undergo a rapid decay in force available for orthodontic tooth movement over time as shown by their performance in Figure #1. This is to be contrasted with that of all of the nickel - titanium springs over time.

Evaluating the results in terms of percentage of force lost over time, it can be said that the lightest of the nickel-titanium springs (GAC 100 g, Masel 100 g) were the most consistent, yielding forces at the end of the twelve weeks which were close to their initial values. The heaviest nickel-titanium spring (GAC 300 g) showed a force decline over the twelve weeks of about 33 % whereas the stainless steel springs lost on average about 42 % of their initial force over just the first four weeks and required reactivation. The other nickel-titanium springs were between the 100 g and 300 g NiTi in terms of performance.

DISCUSSION

The alloy of nickel and titanium known as Nitinol was developed in the early 1960's by William F. Buehler at the Naval Ordnance Laboratory in the U.S.A. (hence its later orthodontic trade name, **Nitinol**).[20] It was originally intended for use in the space program, but after its remarkable properties were made known, it was marketed for orthodontic use . As originally developed, the alloy was a bimetallic compound of 55% nickel and 45% titanium. It exhibited exceptional springback and a flat load/deflection curve. In this form, the alloy consisted of its stabilized martensitic phase. It is known that the alloy has the capacity for super-elasticity and shape-memory i.e. at elevated temperatures the alloy is in its austenitic phase and can be formed. If it is then cooled, it can be deformed but will regain its original shape when reheated. This property was not utilized by the first stabilized work-hardened martensitic form of the alloy as the austenite transition temperature was too elevated. Newer alloys developed in the mid-1970's (Japanese and Chinese NiTi)[4, 12, 14, 17] incorporated active austenitic grain structure i.e. at clinically useful temperatures, there occurs a transition from a predominantly martensitic to a predominantly austenitic phase. They exhibited true 'super-elasticity' with large reversible strains and a non-elastic stress-strain curve delivering essentially constant force over a wide range of deflection. Another unusual feature of this alloy is that its unloading curve varies depending on the degree of initial activation leading to the unique clinical application of re-activating the wire just by releasing it from the brackets and retying it.[17] This 'A- NiTi' is difficult to form but can be shaped by passing electric current through it. It seems to be most useful when large deflections and constant force are required e.g initial arch wires and coil springs.[1]

The behaviour of springs made of various steel alloys is well known and follows a pattern which can be predicted by a formula:

Force delivered = spring extension X constant

with the constant varying for different materials and spring dimensions. Thus it is to be expected that stainless steel springs would deliver progressively less force over time as their extension decreased due to tooth movement. Similarly, the behaviour of elastomeric materials could be represented by a similar formula, were it not for the effects of stress relaxation and material degradation over time. Nonetheless, previous researchers have shown unequivocally forces generated by elastomers do degrade over time in general delivering only half of their initial force value after a four week period.[21-26]. It appears 'super - elastic' NiTi (Japanese, Chinese NiTi) does not offer force degradation that can be predicted by any known formulae[14] offering as it does a relatively flat load/deflection curve. The attraction of nickel-titanium coil springs as a potential constant, predictable force delivery modality is obvious.

The question of what should be the most desirable force magnitude to be applied to teeth to induce orthodontic tooth movement has been addressed in detail by the studies cited in the literature review[27 - 48]. Overall, the review done by Quinn and Yoshikawa[48] neatly sums up all pertinent literature stating that the best available estimate of maximally efficient canine retraction force based on clinical data is 100 - 200 g, equating to a compressive stress on the root of 70 - 140 g/cm^2. They recommend to minimize anchorage loss an appliance should be used which delivers constant forces in this range. It is understood, however, individual variability in such things as root length, bone density and metabolism, age, etc. will cause there to be a range of clinically useful force.

This experiment follows on earlier work by Angolkar et al[8] which demonstrated equivocal results when comparing the force degradation of NiTi to stainless steel over time. In fact, their results showed certain types of NiTi springs lost more force than comparable springs of stainless steel and Elgiloy. However, their study did not if one is using sliding mechanics to retract a tooth, one must apply approximately twice as much force as is ideally needed to induce orthodontic tooth movement in that tooth[61]. Kapila et al[54] have shown one must overcome approximately 193 g of frictional resistance to slide a .019 X .025 inch stainless steel wire through a stainless steel bracket with a slot of .022 inch width. If we use the high end of the ideal force range proposed by Quinn and Yoshikawa[48] and add to it the frictional force (200 g) to be overcome, we can see it is possible a mean force of about 400 g applied constantly may be desirable if one is using .019 X .025 inch wire in a .022 slot with sliding mechanics. That is not to say that such a force would be the best in every clinical situation, only it perhaps is a generic 'ideal' level to strive for when using this wire/bracket combination.

This author could find only one clinical study comparing the space - closure performance of NiTi springs to, in this case, an elastomeric module[27]. Although their sample size was small (n = 17) Samuels et al showed the rate of closure was faster with a 150 g NiTi spring than with an elastomeric module delivering an initial force of 400 - 450 g. Referring to the fears of Bennett and McLaughlin[3] that NiTi springs may be 'too efficient', they stated they did not see any loss of torque or rotational control or excessive tissue pile up in the areas where NiTi springs were placed. They alluded to the potential clinical use of this perceived difference in closure rates between elastomers and NiTi springs in the correction of mid-line discrepancies i.e. use of a NiTi spring on the side towards which the midline has to move.

Cost certainly must be considered as well, as one must ask whether an increase in space closure rate of perhaps 0.5 mm per month warrants the additional cost of such a material. At 1994 prices (drawn from the Masel 1994 catalogue[67]), and if one assumes a four bicuspid extraction case requires about 30 cm of elastomeric chain to close spaces and hold spaces closed, it costs about $3 U.S to close space. On the other hand, the cost per case for

NiTi springs is about $16 U.S. if one uses four springs per patient. The cost drops if one recycles the springs, but it is evident they would have to be reused four or five times to be as cost effective as elastomerics. Although there is evidence that clinical recycling may not necessarily degrade the properties of NiTi wire[11, 13], it is probable there is a maximum number of times this material can be recycled given the extreme conditions it is subjected to in the oral cavity. In fact, it was noted during this study some NiTi springs (Masel 200 g) seemed to undergo a slow deformation by losing their initial symmetrical shape although this did not seem to hamper their performance (see photographs #7 and #8) .

It would seem desirable to follow up this study with a clinical study along the lines of that done by Samuels et al[27] to determine whether the superior force degradation characteristics of nickel-titanium closed-coil springs relative to stainless steel springs and elastomers translates into improved clinical efficiency in space closure.

A potential criticism of the present study is the failure to control temperature effects in a precise manner. It would be more desirable perhaps to leave the springs at a constant temperature throughout the entire experiment, especially during force measurement. It is possible the superior performance of the NiTi springs relative to the stainless steel ones would be further enhanced by controlling this variable.

TABLE #1

SPRING GROUP#	MANUFACTURER	MATERIAL	FORCE AS LABELED
1	TP Orthodontics	NiTi	150g
2	TP Orthodontics	NiTi	250g
3	Rocky Mountain	NiTi	200g
4	Masel	NiTi	200g
5	Masel	NiTi	100g
6	Unitek	HI-T Stainless Steel	Not Given
7	GAC	NiTi	100g
8	GAC	NiTi	300g
9	Unitek	HI-T Stainless Steel	Not Given
10	GAC	NiTi	200g

TABLE #2

Comparisons of Spring Means, Scheffe' Test

	M 100	M 200	TP150	G 100	TP250	RMO	G 200	U 9	G 300	U 12
Means:	79.5	108.5	109.7	113.7	127.3	183.2	236.6	283.8	398.7	681.0
M 100	-					***	***	***	***	***
M 200		-				**	***	***	***	***
TP150			-			**	***	***	***	***
G 100				-		*	***	***	***	***
TP250					-		***	***	***	***
RMO						-		***	***	***
G 200							-		***	***
U 9								-	***	***
G 300									-	***
U 12										-

* P<.05   ** P<.01   *** P<.001

M 100 = Masel 100 g NiTi
M 200 = Masel 200 g NiTi
TP 250 = TP 250 g NiTi
G 100 = GAC 100 g NiTi
TP 250 = TP 250 g NiTi
RMO = RMO 200 g NiTi
G 200 = GAC 200 g NiTi
U 9 = Unitek 9 X 30 stainless steel
G 300 = GAC 300 g NiTi
U 12 = Unitek 12 X 30 stainless steel

TABLE #3

Analysis of Simple Effects

Spring	F
GAC 100gm	0.23
Masel 100gm NIT	0.29
RMO	2.57**
TP 150gm NIT	3.05**
GAC 200gm	3.24**
Masel 200gm NIT	4.18**
TP 250gm NIT	11.67**
Unitek 9x20 SS	61.38**
GAC 300gm	64.57**
Unitek 12x30 SS	511.49**

Figure #1: Force Changes Over Time

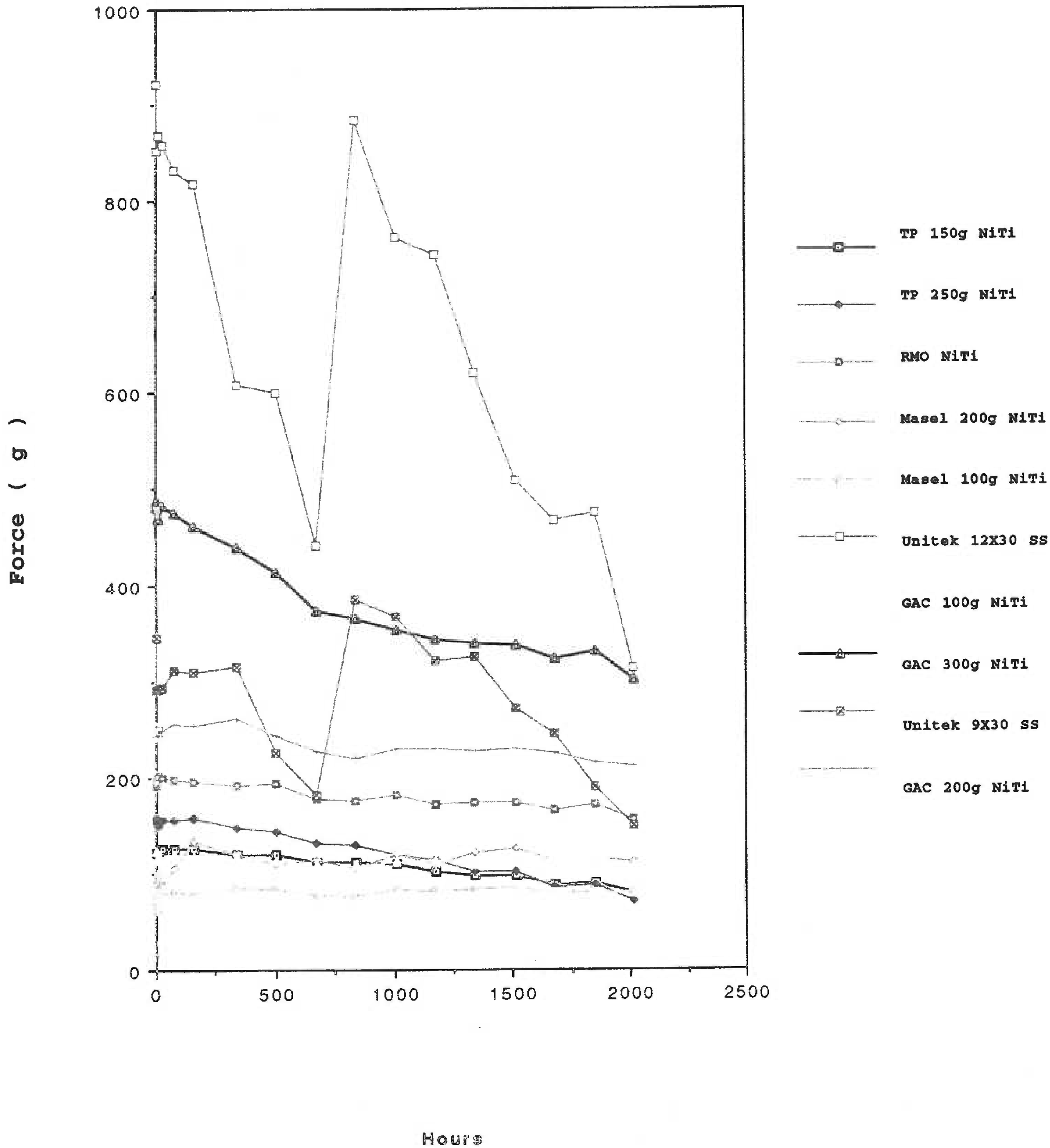


Figure #2: Masel 100g Changes Over Time

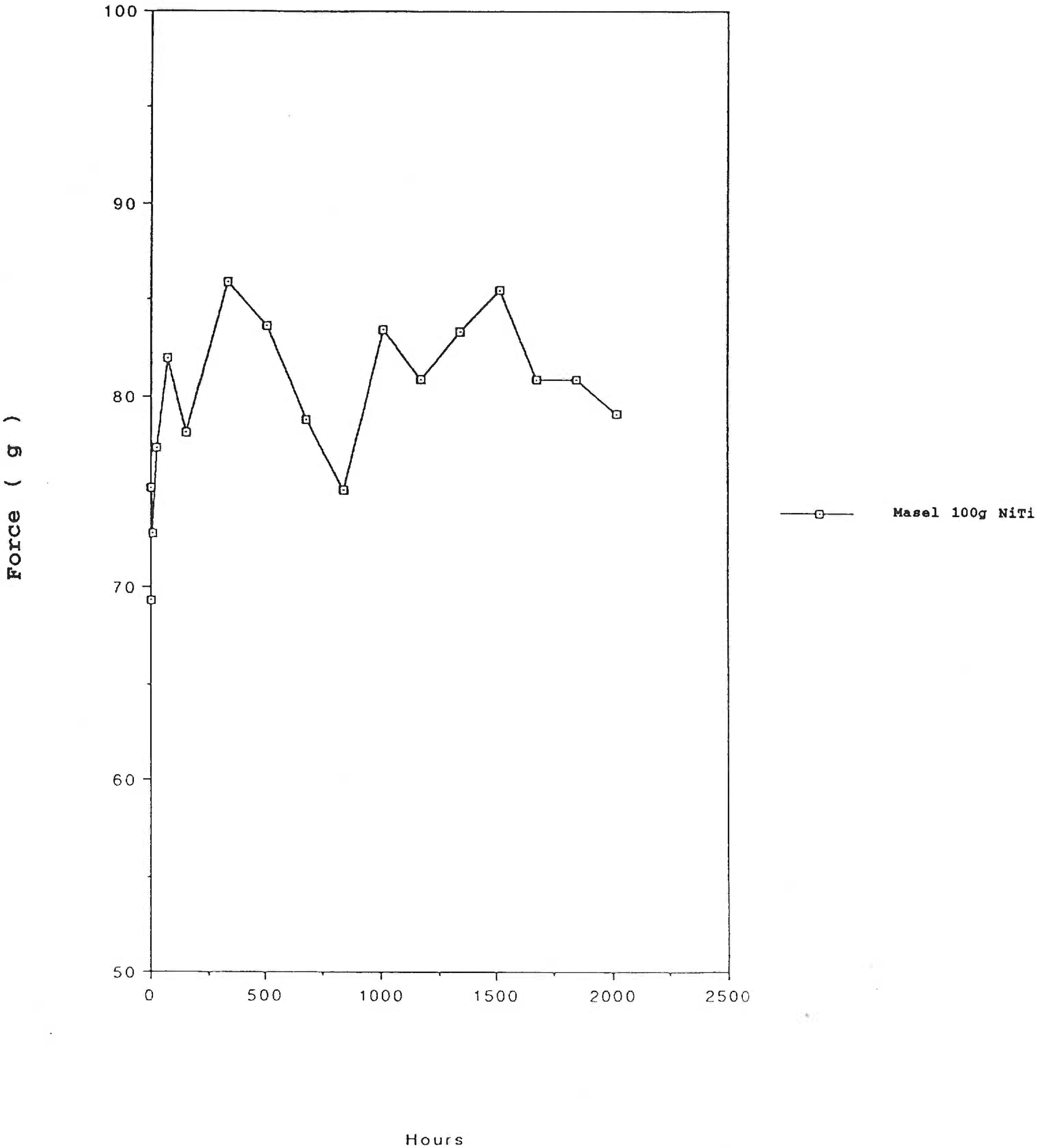


Figure #3: GAC 100g Changes Over Time

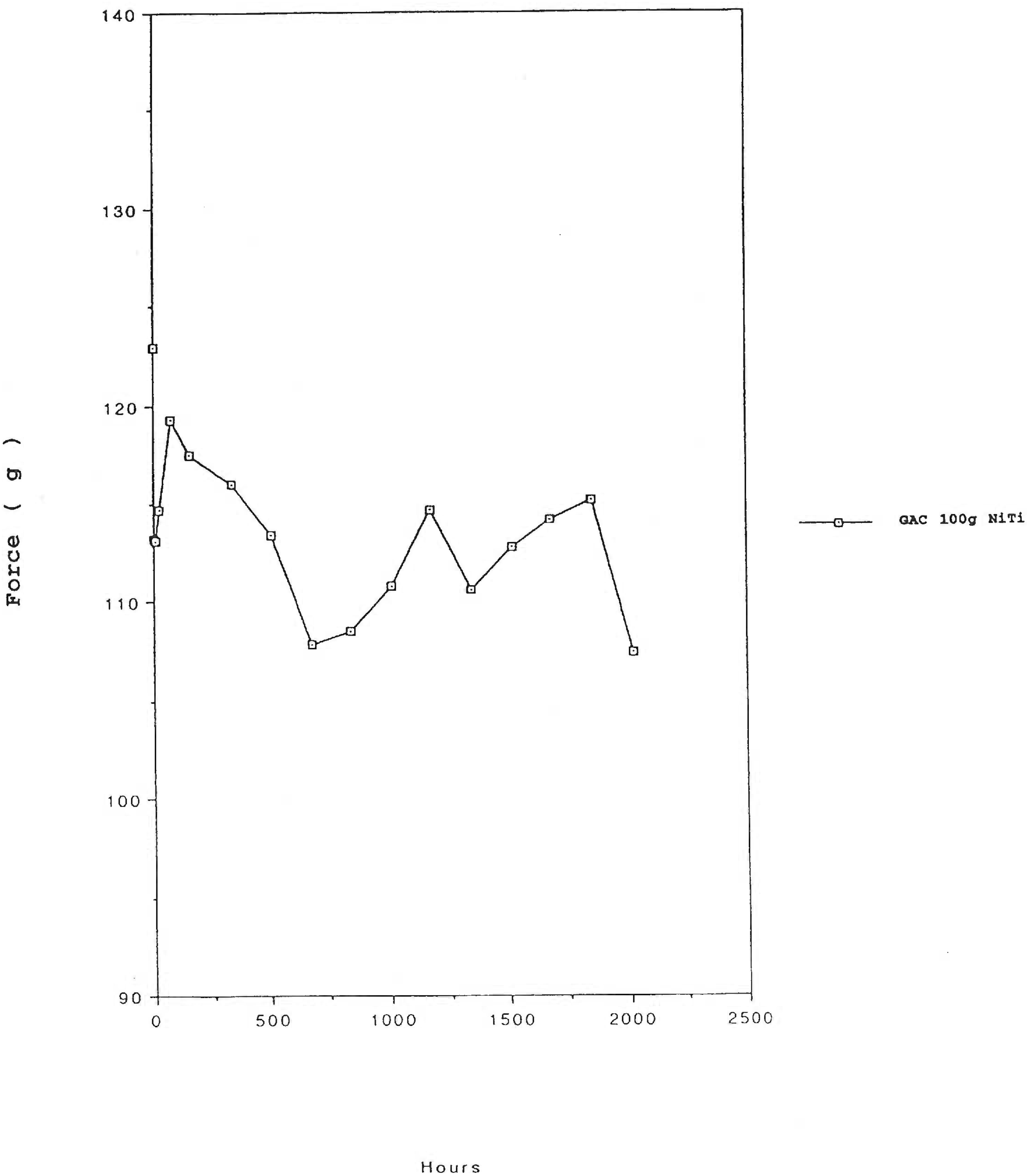
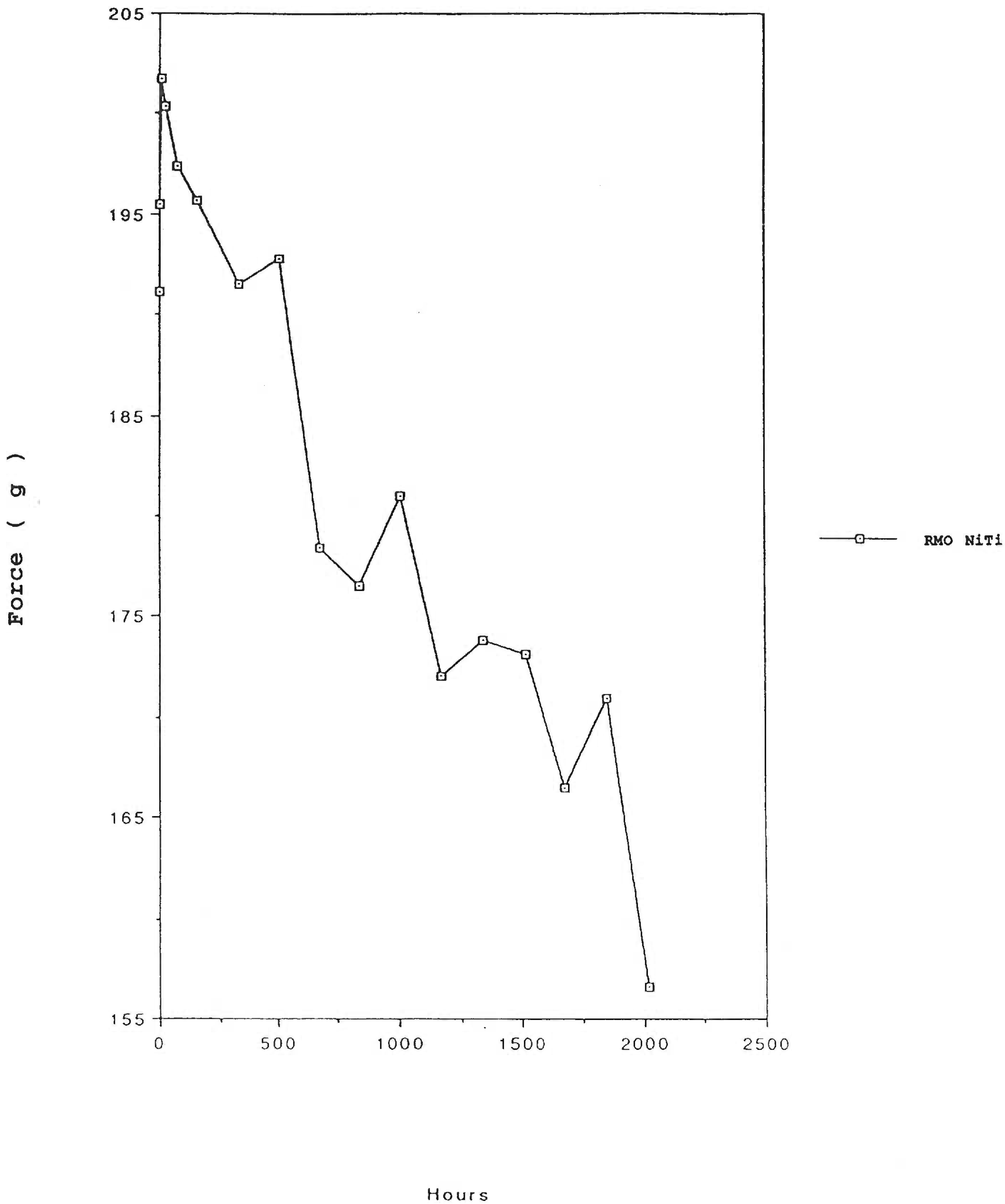
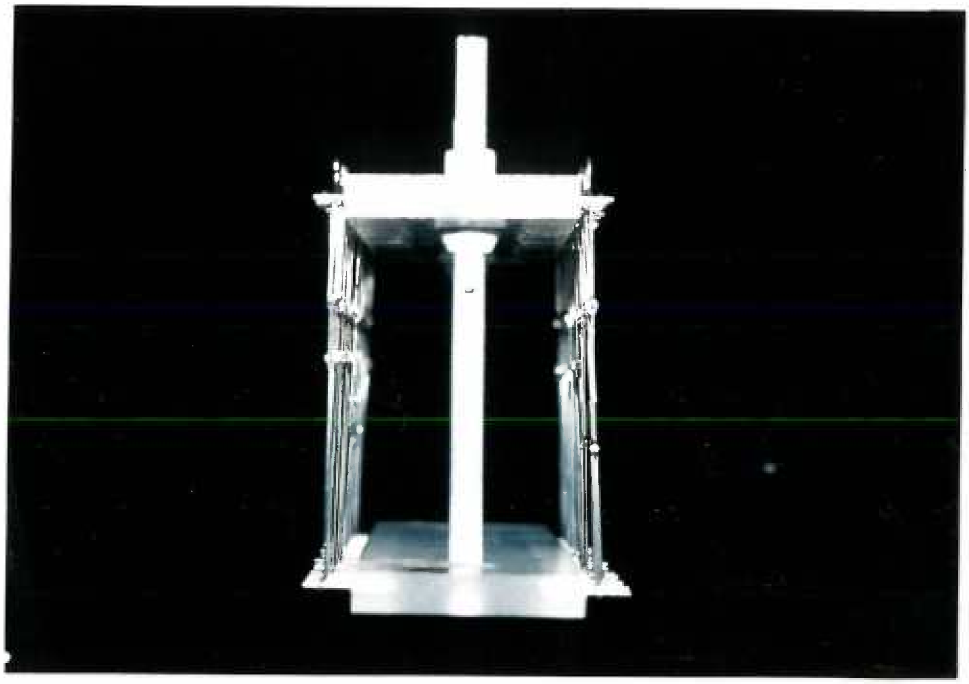
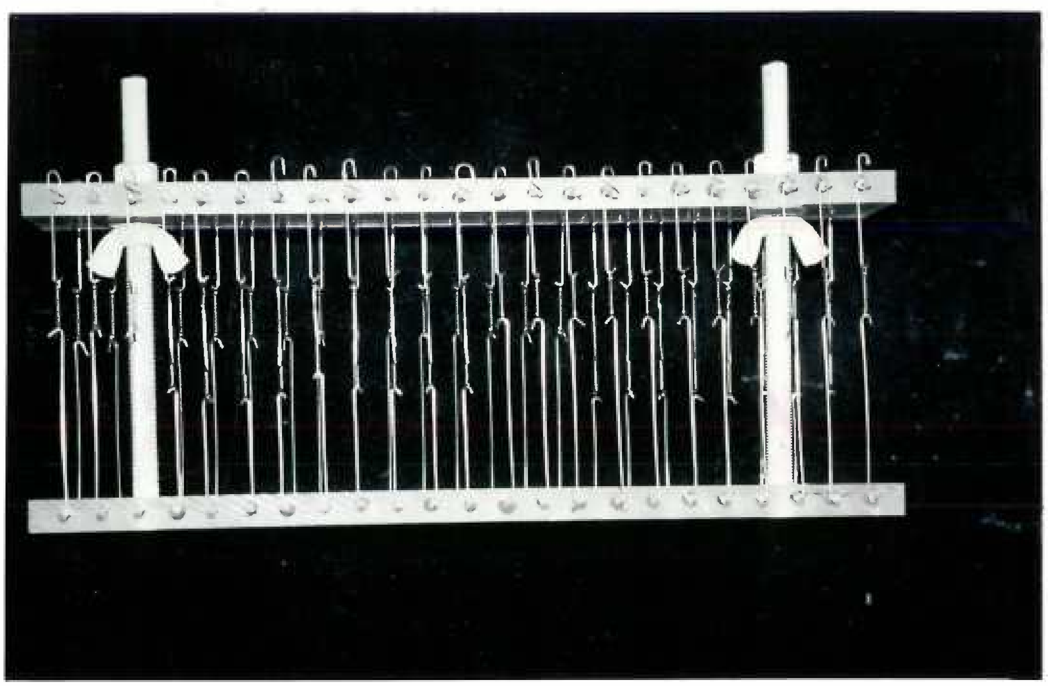


Figure #4: RMO Changes Over Time

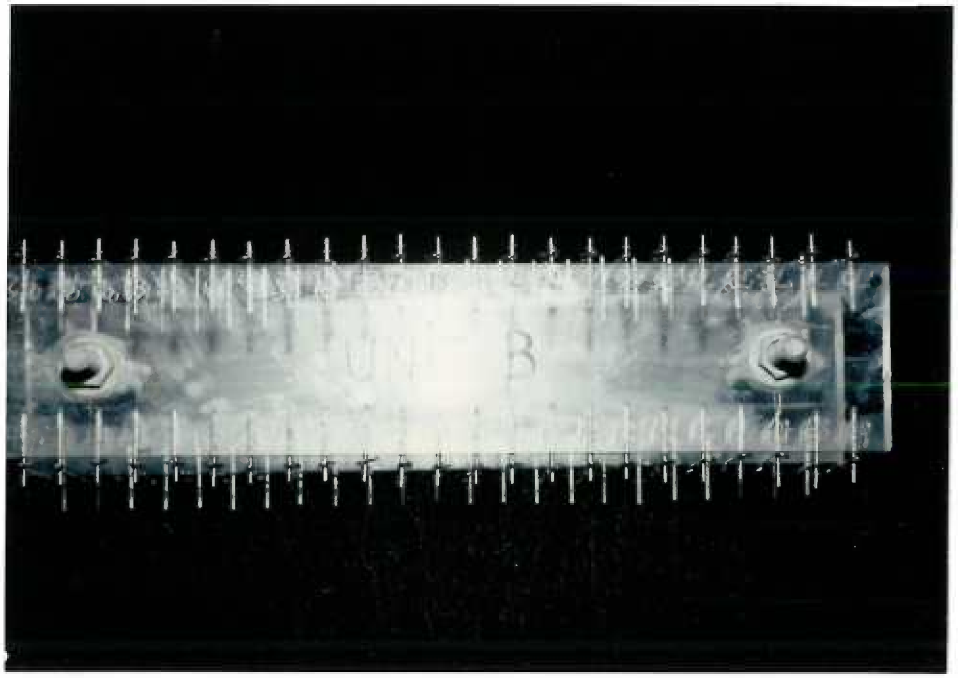




Photograph #1. Experimental apparatus, end - on view.

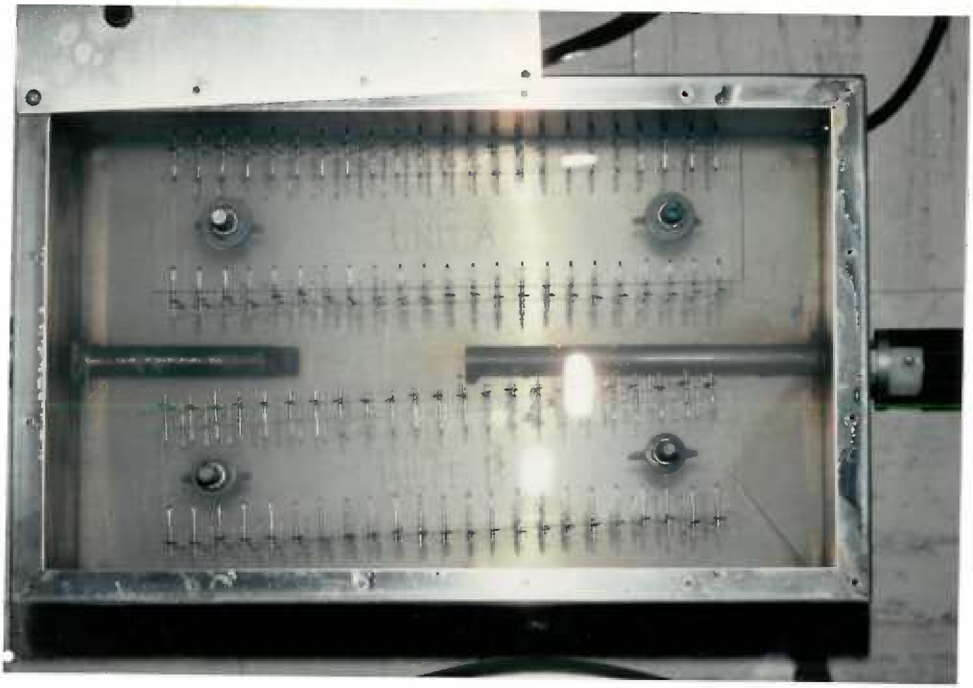


Photograph #2. Experimental apparatus, lateral view.



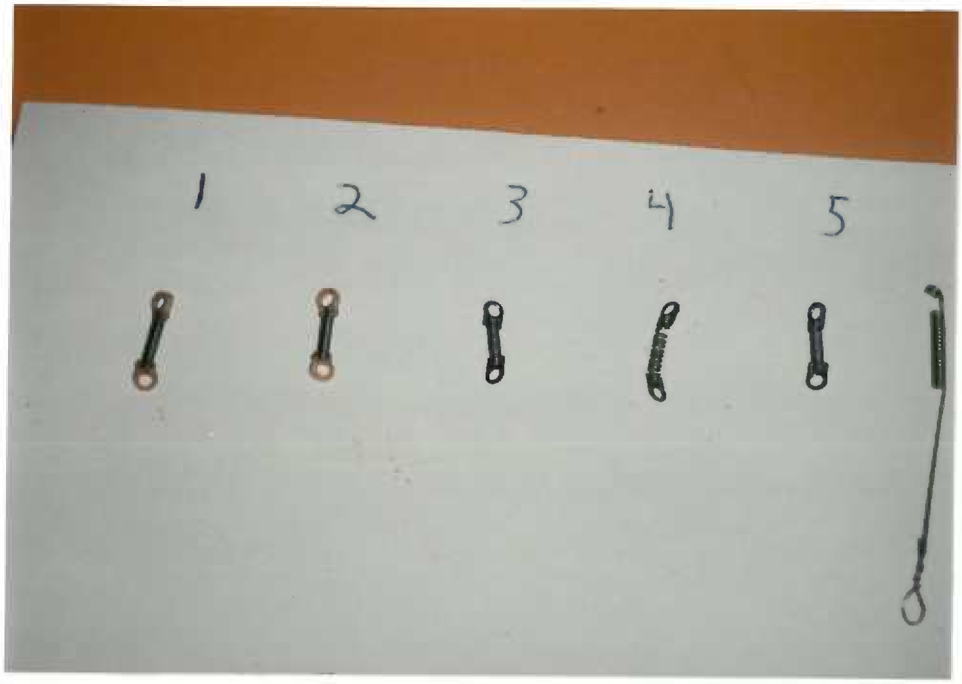
Photograph #3. Experimental apparatus, view from above.

Photograph #4. Controlled - temperature water bath.

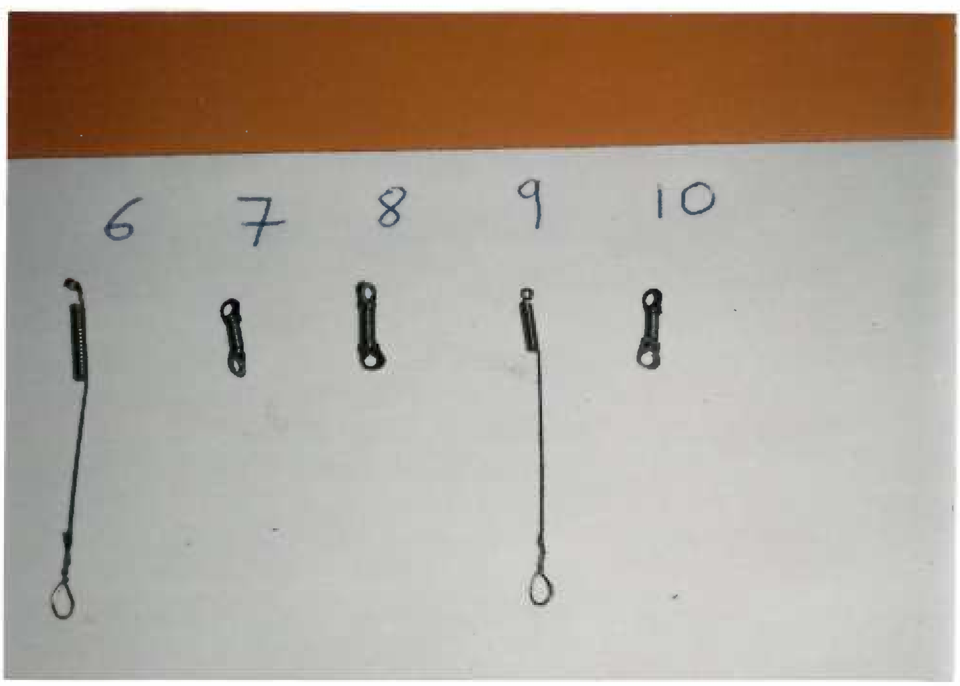


Photograph #5. Experimental jigs immersed in water bath.

Photograph #6. Accu - Force Cadet digital gauge (in case).



Photograph #7. Spring condition at end of experiment. Numbers correspond to legend in Table #1, pp. 34. Note spring #4.



Photograph #8. Spring condition at end of experiment. Numbers correspond to legend in Table #1, pp. 34.

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APPENDIX

	Springs	0 Hours	0hr-2	1 Hour	1hr-2
1	TP 150gmNIT	106	101	111	112
2	TP 150gmNIT	129	127	124	132
3	TP 150gmNIT	124	129	124	126
4	TP 150gmNIT	127	131	126	121
5	TP 150gmNIT	113	126	122	121
6	TP 150gmNIT	131	127	122	124
7	TP 150gmNIT	114	124	119	117
8	TP 150gmNIT	122	121	119	127
9	TP 150gmNIT	119	121	121	126
10	TP 250gm NIT	146	147	144	141
11	TP 250gm NIT	152	151	144	147
12	TP 250gm NIT	151	151	141	146
13	TP 250gm NIT	167	162	157	154
14	TP 250gm NIT	156	156	151	154
15	TP 250gm NIT	146	162	159	159
16	TP 250gm NIT	169	171	161	164
17	TP 250gm NIT	159	156	154	156
18	TP 250gm NIT	161	162	156	166
19	RMO	196	201	202	207
20	RMO	201	202	206	204
21	RMO	132	131	139	132
22	RMO	204	197	215	212
23	RMO	255	216	240	235
24	RMO	186	186	182	195
25	RMO	196	189	196	191
26	RMO	197	196	197	192
27	RMO	179	176	184	189
28	Masel 200gm ...	75	81	59	63
29	Masel 200gm ...	110	99	56	63
30	Masel 200gm ...	86	78	63	69
31	Masel 200gm ...	73	79	43	39
32	Masel 200gm ...	106	99	61	66
33	Masel 200gm ...	103	98	61	66
34	Masel 200gm ...	93	81	54	65
35	Masel 200gm ...	83	81	71	65
36	Masel 200gm ...	94	88	63	73
37	Masel 100gm ...	56	64	61	68
38	Masel 100gm ...	63	63	68	68
39	Masel 100gm ...	71	69	63	64
40	Masel 100gm ...	78	78	74	71
41	Masel 100gm ...	69	68	74	83
42	Masel 100gm ...	79	73	81	81
43	Masel 100gm ...	68	68	76	79
44	Masel 100gm ...	71	69	84	83
45	Masel 100gm ...	71	71	91	84
46	Unitek 12x30 ...	908	918	1020	1028
47	Unitek 12x30 ...	923	913	877	898
48	Unitek 12x30 ...	1073	1065	995	1009
49	Unitek 12x30 ...	907	872	795	785

Table A

Raw data across time, with forces measured in grams. Two measures were taken at each time interval.

	8 Hours	8hr-2	24 Hours	24hr-2	3 Days
1	111	106	116	116	111
2	124	122	132	127	129
3	122	124	126	126	126
4	121	119	129	127	127
5	119	117	127	122	127
6	122	122	127	121	129
7	121	114	124	122	126
8	117	121	127	127	127
9	111	117	124	126	127
10	137	136	151	146	147
11	144	147	147	151	149
12	139	142	152	147	149
13	156	154	162	157	164
14	144	149	159	152	156
15	151	152	157	157	159
16	161	159	161	162	164
17	151	149	156	156	156
18	159	161	159	161	161
19	204	194	201	202	204
20	207	201	207	212	206
21	139	131	139	137	136
22	215	219	214	211	214
23	245	255	229	232	232
24	209	203	206	204	184
25	207	201	204	209	204
26	207	207	207	202	197
27	191	196	196	194	194
28	114	106	99	99	93
29	131	112	88	98	106
30	116	119	104	114	112
31	84	89	83	69	83
32	119	119	98	106	119
33	104	99	86	78	131
34	83	78	71	78	93
35	73	84	84	74	121
36	89	72	71	71	91
37	73	71	73	73	88
38	68	66	73	76	79
39	69	68	74	78	83
40	69	69	73	74	84
41	69	71	73	71	81
42	76	74	81	79	79
43	74	66	76	76	79
44	79	76	84	83	83
45	86	86	86	88	86
46	975	939	963	943	898
47	888	890	890	895	842
48	1028	1018	1010	1030	940
49	822	817	849	900	799

Table A (continued)

	8 Hours	8hr-2	24 Hours	24hr-2	3 Days
50	897	923	872	875	840
51	870	894	839	852	858
52	868	830	809	799	792
53	711	708	706	690	706
54	769	777	750	768	805
55	114	117	119	117	116
56	111	107	106	107	114
57	122	117	124	119	124
58	103	101	107	111	111
59	101	104	104	106	109
60	116	122	129	119	129
61	116	122	119	119	126
62	106	104	107	107	119
63	127	126	126	119	129
64	446	466	478	471	473
65	425	427	456	450	446
66	476	471	486	485	486
67	463	475	475	476	475
68	511	520	518	518	503
69	539	550	568	579	533
70	473	483	491	495	488
71	427	428	443	440	431
72	436	430	441	440	435
73	397	380	377	375	377
74	297	303	290	295	310
75	264	272	272	274	279
76	353	348	340	357	362
77	320	320	338	320	352
78	284	277	292	290	314
79	285	282	279	288	317
80	232	211	224	225	239
81	222	215	225	219	245
82	225	232	237	242	250
83	224	222	229	225	239
84	212	215	224	220	229
85	237	235	247	250	255
86	240	244	244	249	255
87	254	252	262	255	264
88	252	257	259	264	267
89	279	279	279	280	282
90	264	269	269	269	259
91	.	.	.	.	.

Table A (continued)

	3days-2	1 Week	1wk-2	2 Weeks	2wk-2
1	121	107	109	99	103
2	127	124	122	117	116
3	126	121	121	119	119
4	131	126	126	124	117
5	126	126	127	124	121
6	127	129	127	129	124
7	131	136	125	119	124
8	131	136	131	127	131
9	127	131	127	122	126
10	144	147	146	141	137
11	151	157	156	144	144
12	149	154	151	141	139
13	162	171	162	157	157
14	159	156	157	149	147
15	159	159	159	151	152
16	162	167	164	154	156
17	157	152	152	144	146
18	161	164	162	154	151
19	209	202	199	204	206
20	207	204	199	199	194
21	137	137	134	131	129
22	214	202	206	196	199
23	230	219	219	206	204
24	187	204	197	194	196
25	209	204	201	201	199
26	194	202	201	199	207
27	194	196	197	189	194
28	96	139	126	109	98
29	100	141	132	114	106
30	107	137	153	126	126
31	88	106	112	104	112
32	127	146	138	126	141
33	121	147	156	136	137
34	91	132	122	117	116
35	114	139	132	126	117
36	98	136	118	132	108
37	79	78	79	83	86
38	78	79	78	88	83
39	89	86	84	88	83
40	79	76	79	86	91
41	79	79	79	81	81
42	83	79	76	91	88
43	76	73	74	76	79
44	84	74	76	89	89
45	86	73	83	91	93
46	890	812	822	844	824
47	832	860	870	603	594
48	970	975	948	699	694
49	805	797	797	573	545

Table A (continued)

	3days-2	1 Week	1wk-2	2 Weeks	2wk-2
50	845	820	830	621	609
51	855	809	824	594	593
52	797	785	775	556	580
53	706	716	721	473	470
54	791	772	780	543	560
55	121	116	112	116	116
56	116	111	112	106	107
57	117	129	121	131	124
58	112	111	112	114	109
59	112	111	107	111	107
60	126	124	122	121	121
61	124	124	122	119	119
62	114	114	114	111	107
63	127	127	126	126	122
64	480	466	473	420	442
65	450	443	453	405	408
66	485	466	475	422	427
67	483	461	455	418	422
68	500	485	486	428	422
69	536	475	496	446	431
70	493	456	468	528	492
71	421	423	430	458	453
72	434	430	442	450	448
73	377	382	385	382	390
74	300	304	307	320	314
75	282	282	282	297	292
76	363	367	360	368	365
77	348	345	342	368	370
78	312	310	304	315	310
79	322	330	314	322	315
80	245	239	230	244	239
81	256	254	249	237	242
82	250	242	244	257	259
83	245	229	234	239	245
84	232	232	227	237	244
85	254	257	252	262	252
86	255	250	254	260	252
87	262	259	260	274	269
88	274	255	262	270	265
89	279	284	280	285	285
90	259	265	269	272	282
91	.	.	.	.	.

Table A (continued)

	3 Weeks	3 Weeks-2	4 Weeks	4 Wk - 2	5 Weeks
1	106	103	94	96	96
2	116	119	111	111	111
3	121	121	112	112	111
4	121	117	114	114	117
5	124	119	116	114	114
6	117	116	112	117	114
7	119	124	117	117	111
8	129	124	119	119	116
9	126	119	112	112	112
10	134	136	126	131	124
11	141	141	126	132	132
12	137	132	122	124	121
13	152	149	142	141	139
14	139	144	129	134	136
15	146	146	131	129	127
16	152	149	132	139	134
17	142	139	127	122	127
18	146	147	136	136	131
19	199	192	184	184	179
20	194	191	172	174	174
21	132	134	122	126	121
22	202	192	172	171	186
23	206	209	186	187	191
24	204	211	194	184	187
25	202	199	189	191	184
26	199	204	192	192	184
27	204	197	196	196	179
28	91	98	107	99	101
29	112	106	111	111	98
30	106	107	104	109	109
31	99	106	109	98	88
32	134	132	134	136	126
33	114	121	124	129	122
34	107	121	124	108	111
35	104	109	119	101	107
36	112	103	111	109	106
37	84	84	81	86	79
38	79	84	79	79	73
39	86	89	81	81	78
40	83	84	78	83	69
41	84	81	76	79	74
42	84	86	79	76	81
43	81	78	74	74	71
44	84	89	74	76	78
45	83	83	78	83	73
46	701	730	589	583	842
47	647	654	418	427	1168
48	739	729	495	487	1033
49	569	564	367	358	800

Table A (continued)

	3 Weeks	3 Weeks-2	4 Weeks	4 Wk - 2	5 Weeks
50	598	618	473	473	749
51	583	574	441	445	792
52	564	559	427	423	844
53	465	465	342	347	812
54	533	528	417	412	920
55	116	114	107	107	106
56	107	109	103	101	101
57	114	121	124	112	112
58	112	109	104	104	103
59	106	109	101	103	104
60	119	117	112	111	116
61	112	119	109	112	114
62	111	112	104	106	107
63	116	119	112	109	114
64	410	418	370	375	377
65	397	383	355	365	348
66	410	420	383	380	365
67	398	402	363	376	363
68	423	423	383	380	390
69	431	440	373	380	363
70	431	441	377	382	378
71	412	408	370	368	355
72	402	403	377	367	352
73	295	302	240	242	330
74	227	234	186	176	418
75	196	197	159	164	322
76	284	279	230	239	427
77	254	262	219	224	393
78	227	222	182	187	377
79	230	237	187	197	412
80	146	149	104	106	385
81	156	164	122	121	380
82	239	234	225	229	220
83	227	229	220	219	212
84	219	225	214	206	197
85	242	244	225	232	222
86	239	240	230	229	222
87	245	249	230	232	224
88	255	254	225	222	224
89	264	265	244	244	230
90	250	257	229	225	220
91	.	.	.	.	.

Table A (continued)

	5 Wk - 2	6 Weeks	6 Wk - 2	7 Weeks	7 Wk - 2
1	94	96	86	86	84
2	107	109	106	99	98
3	112	111	111	104	104
4	117	112	112	109	104
5	109	111	112	106	109
6	111	112	107	107	99
7	116	116	114	106	109
8	121	111	119	109	107
9	116	112	109	101	103
10	126	119	116	111	111
11	127	119	124	111	107
12	121	112	114	101	106
13	139	132	132	124	127
14	131	121	119	117	116
15	131	121	116	114	112
16	134	121	126	116	117
17	124	116	112	107	103
18	131	126	126	119	117
19	179	192	192	186	182
20	176	172	166	167	166
21	124	124	127	117	122
22	181	187	189	174	177
23	189	201	202	191	189
24	191	197	199	176	179
25	184	186	187	176	176
26	181	187	182	174	177
27	187	187	181	187	181
28	91	116	112	103	104
29	99	116	112	104	103
30	89	117	119	88	83
31	94	96	93	86	96
32	122	147	146	144	144
33	124	131	134	122	131
34	101	124	116	111	101
35	104	121	122	106	116
36	107	124	124	121	116
37	76	91	86	88	86
38	73	83	84	78	79
39	76	86	84	78	81
40	73	88	79	86	81
41	79	84	79	81	83
42	71	76	84	73	76
43	71	74	79	79	76
44	78	84	81	81	79
45	79	86	93	83	88
46	835	800	809	769	793
47	1176	1018	1021	988	977
48	1035	877	867	878	870
49	825	712	693	707	679

Table A (continued)

	5 Wk - 2	6 Weeks	6 Wk - 2	7 Weeks	7 Wk - 2
50	711	596	619	573	583
51	802	623	676	654	646
52	823	737	736	707	696
53	835	696	714	696	687
54	904	770	763	749	763
55	111	109	111	121	121
56	99	109	112	111	111
57	114	116	114	121	121
58	103	107	109	116	109
59	107	109	107	109	112
60	119	119	117	116	119
61	109	112	109	114	119
62	104	107	107	109	107
63	111	109	112	114	114
64	383	355	362	368	353
65	353	333	330	342	342
66	372	350	357	352	352
67	367	355	353	335	335
68	374	358	358	347	345
69	357	362	341	348	350
70	370	360	370	343	335
71	363	350	350	328	335
72	358	357	345	330	346
73	330	322	317	262	267
74	427	405	408	348	340
75	341	307	319	254	259
76	445	390	395	370	363
77	387	382	375	343	323
78	378	352	364	332	313
79	418	387	403	362	363
80	393	373	379	332	324
81	383	355	365	327	320
82	219	232	230	230	229
83	217	217	214	219	225
84	204	211	211	214	212
85	229	232	237	230	232
86	222	232	230	234	230
87	229	232	240	235	229
88	217	234	232	237	237
89	237	242	244	242	242
90	224	227	220	234	223

Table A (continued)

	8 Weeks	8 Wks - 2	9 Weeks	9 Wks - 2	10 Weeks
1	73	74	78	79	63
2	96	94	98	99	89
3	103	101	103	101	93
4	101	101	101	99	89
5	101	99	101	103	96
6	107	103	96	99	91
7	104	106	104	96	91
8	99	101	103	103	86
9	96	98	96	96	84
10	104	103	106	104	84
11	106	104	103	101	86
12	91	91	94	94	76
13	111	112	114	111	94
14	104	98	106	101	81
15	99	93	96	101	83
16	106	104	104	106	91
17	99	96	93	99	89
18	106	116	106	106	89
19	177	182	196	182	166
20	157	161	162	156	151
21	122	126	124	122	117
22	177	181	171	174	164
23	192	196	189	194	187
24	187	184	192	192	191
25	182	184	184	181	189
26	174	177	174	160	166
27	189	181	181	182	186
28	109	123	124	132	107
29	106	122	132	125	119
30	101	98	111	101	101
31	111	100	103	98	96
32	149	147	157	156	137
33	141	142	144	138	134
34	127	121	119	122	104
35	134	124	126	134	119
36	127	121	124	111	119
37	88	83	93	91	86
38	83	83	84	84	83
39	83	81	91	91	91
40	86	81	83	83	86
41	84	86	86	86	79
42	84	79	86	78	78
43	79	76	78	73	74
44	89	83	86	89	81
45	88	84	89	88	81
46	755	750	505	558	480
47	897	907	742	766	654
48	769	742	694	671	647
49	541	526	392	400	418

Table A (continued)

	8 Weeks	8 Wks - 2	9 Weeks	9 Wks - 2	10 Weeks
50	453	438	377	360	307
51	511	536	418	419	370
52	546	551	463	463	441
53	528	500	445	472	441
54	624	594	493	520	453
55	109	111	104	111	121
56	106	107	107	111	117
57	116	116	116	122	119
58	107	112	116	112	109
59	107	106	107	112	107
60	112	117	119	121	119
61	117	111	114	116	116
62	107	107	107	109	104
63	112	111	112	114	117
64	352	357	357	342	347
65	337	333	323	328	319
66	337	342	345	335	332
67	337	329	323	317	317
68	327	337	320	323	319
69	320	326	322	312	310
70	370	353	360	355	342
71	328	343	342	344	328
72	330	344	353	358	320
73	264	276	235	242	184
74	352	347	297	297	255
75	279	285	196	202	197
76	367	385	312	330	304
77	320	323	272	274	259
78	319	319	270	278	237
79	360	362	315	314	267
80	320	345	269	257	247
81	302	322	252	259	242
82	232	222	230	232	232
83	220	222	215	224	222
84	214	209	209	214	211
85	229	229	237	225	224
86	230	227	234	223	235
87	232	229	235	237	234
88	235	237	235	237	229
89	247	242	239	245	240
90	224	232	220	225	224

Table A (continued)

	10 Wks - 2	11 Weeks	11 Wks - 2	12 Weeks	12 Wks - 2
1	68	71	68	59	56
2	84	91	91	84	84
3	94	104	99	83	86
4	88	96	96	86	84
5	89	96	96	84	91
6	98	88	93	89	81
7	96	98	91	78	79
8	89	88	88	83	84
9	86	84	84	78	74
10	91	88	91	79	81
11	86	91	96	76	74
12	78	78	86	59	58
13	98	98	91	83	83
14	88	88	81	68	64
15	84	86	81	73	68
16	89	89	94	74	78
17	79	76	79	68	61
18	89	89	93	73	69
19	164	177	177	161	161
20	144	149	151	137	137
21	114	119	119	103	106
22	164	172	181	156	151
23	174	191	186	192	181
24	191	192	192	179	176
25	182	184	186	174	181
26	167	164	169	147	151
27	181	189	179	162	164
28	111	114	106	106	103
29	121	137	94	126	116
30	99	107	101	89	99
31	89	84	76	83	93
32	136	142	132	127	136
33	126	136	151	127	131
34	104	101	114	107	109
35	127	146	109	111	107
36	109	117	117	117	122
37	84	81	76	79	78
38	74	79	78	78	76
39	81	83	83	83	81
40	88	84	88	79	81
41	83	81	81	74	79
42	76	88	78	76	84
43	73	76	73	73	71
44	76	84	83	83	79
45	81	81	78	84	86
46	478	423	427	310	304
47	685	656	646	496	498
48	649	621	628	485	493
49	408	418	475	282	287

Table A (continued)

	10 Wks - 2	11 Weeks	11 Wks - 2	12 Weeks	12 Wks - 2
50	307	373	367	177	186
51	367	388	402	240	229
52	420	480	407	282	333
53	391	433	435	270	279
54	487	491	475	243	253
55	114	117	119	112	107
56	116	114	112	107	104
57	119	121	112	117	114
58	111	114	112	103	101
59	106	117	107	104	106
60	117	119	114	117	109
61	114	122	117	111	109
62	111	117	109	99	103
63	119	119	112	104	107
64	337	358	345	304	314
65	306	338	347	280	287
66	310	327	330	294	302
67	326	343	330	290	292
68	319	345	330	289	287
69	319	320	317	287	289
70	338	350	335	320	315
71	323	323	300	320	314
72	317	307	323	315	312
73	167	126	117	101	94
74	282	229	222	171	154
75	194	176	127	96	104
76	307	240	242	204	197
77	255	194	189	149	156
78	245	189	199	154	146
79	298	202	224	187	192
80	242	171	174	146	149
81	254	201	204	147	151
82	225	222	227	215	214
83	220	207	202	211	207
84	206	202	199	206	199
85	222	211	219	214	220
86	230	220	224	217	217
87	224	217	222	214	212
88	229	222	222	215	217
89	229	224	222	215	222
90	220	219	209	202	202

Table A (continued)

	0	1	8	24	72	168
1	103.5	111.5	108.5	116.0	116.0	108.0
2	128.0	128.0	123.0	129.5	128.0	123.0
3	126.5	125.0	123.0	126.0	126.0	121.0
4	129.0	123.5	120.0	128.0	129.0	126.0
5	119.5	121.5	118.0	124.5	126.5	126.5
6	129.0	123.0	122.0	124.0	128.0	128.0
7	119.0	118.0	117.5	123.0	128.5	130.5
8	121.5	123.0	119.0	127.0	129.0	133.5
9	120.0	123.5	114.0	125.0	127.0	129.0
10	146.5	142.5	136.5	148.5	145.5	146.5
11	151.5	145.5	145.5	149.0	150.0	156.5
12	151.0	143.5	140.5	149.5	149.0	152.5
13	164.5	155.5	155.0	159.5	163.0	166.5
14	156.0	152.5	146.5	155.5	157.5	156.5
15	154.0	159.0	151.5	157.0	159.0	159.0
16	170.0	162.5	160.0	161.5	163.0	165.5
17	157.5	155.0	150.0	156.0	156.5	152.0
18	161.5	161.0	160.0	160.0	161.0	163.0
19	198.5	204.5	199.0	201.5	206.5	200.5
20	201.5	205.0	204.0	209.5	206.5	201.5
21	131.5	135.5	135.0	138.0	136.5	135.5
22	200.5	213.5	217.0	212.5	214.0	204.0
23	235.5	237.5	250.0	230.5	231.0	219.0
24	186.0	188.5	206.0	205.0	185.5	200.5
25	192.5	193.5	204.0	206.5	206.5	202.5
26	196.5	194.5	207.0	204.5	195.5	201.5
27	177.5	186.5	193.5	195.0	194.0	196.5
28	78.0	61.0	110.0	99.0	94.5	132.5
29	104.5	59.5	121.5	93.0	103.0	136.5
30	82.0	66.0	117.5	109.0	109.5	145.0
31	76.0	41.0	86.5	76.0	85.5	109.0
32	102.5	63.5	119.0	102.0	123.0	142.0
33	100.5	63.5	101.5	82.0	126.0	151.5
34	87.0	59.5	80.5	74.5	92.0	127.0
35	82.0	68.0	78.5	79.0	117.5	135.5
36	91.0	68.0	80.5	71.0	94.5	127.0
37	60.0	64.5	72.0	73.0	83.5	78.5
38	63.0	68.0	67.0	74.5	78.5	78.5
39	70.0	63.5	68.5	76.0	86.0	85.0
40	78.0	72.5	69.0	73.5	81.5	77.5
41	68.5	78.5	70.0	72.0	80.0	79.0
42	76.0	81.0	75.0	80.0	81.0	77.5
43	68.0	77.5	70.0	76.0	77.5	73.5
44	70.0	83.5	77.5	83.5	83.5	75.0
45	71.0	87.5	86.0	87.0	86.0	78.0
46	913.0	1024.0	957.0	953.0	894.0	817.0
47	918.0	887.5	889.0	892.5	837.0	865.0
48	1069.0	1002.0	1023.0	1020.0	955.0	961.5
49	889.5	790.0	819.5	874.5	802.0	797.0

Table B.

Mean force measurements, in grams, across time (in hours). Spring numbers correspond to those in Table A.

	0	1	8	24	72	168
50	958.5	885.0	910.0	873.5	842.5	825.0
51	942.5	859.5	882.0	845.5	856.5	816.5
52	901.5	797.0	849.0	804.0	794.5	780.0
53	815.5	679.0	709.5	698.0	706.0	718.5
54	900.0	745.0	773.0	759.0	798.0	776.0
55	119.0	111.5	115.5	118.0	118.5	114.0
56	114.0	103.5	109.0	106.5	115.0	111.5
57	133.0	136.0	119.5	121.5	120.5	125.0
58	111.5	99.0	102.0	109.0	111.5	111.5
59	115.5	101.0	102.5	105.0	110.5	109.0
60	129.5	119.0	119.0	124.0	127.5	123.0
61	130.0	116.5	119.0	119.0	125.0	123.0
62	117.0	109.5	105.0	107.0	116.5	114.0
63	136.5	122.5	126.5	122.5	128.0	126.5
64	494.0	467.0	456.0	474.5	476.5	469.5
65	469.0	441.5	426.0	453.0	448.0	448.0
66	495.5	489.5	473.5	485.5	485.5	470.5
67	474.5	490.5	469.0	475.5	479.0	458.0
68	505.5	534.5	515.5	518.0	501.5	485.5
69	522.0	595.0	544.5	573.5	534.5	485.5
70	487.0	481.5	478.0	493.0	490.5	462.0
71	437.5	446.5	427.5	441.5	426.0	426.5
72	447.0	434.5	433.0	440.5	434.5	436.0
73	412.5	381.5	388.5	376.0	377.0	383.5
74	332.5	304.5	300.0	292.5	305.0	305.5
75	307.5	263.5	268.0	273.0	280.5	282.0
76	394.0	352.5	350.5	348.5	362.5	363.5
77	376.0	326.0	320.0	329.0	350.0	343.5
78	355.5	279.5	280.5	291.0	313.0	307.0
79	370.0	289.0	283.5	283.5	319.5	322.0
80	282.0	202.5	221.5	224.5	242.0	234.5
81	277.0	217.0	218.5	222.0	250.5	251.5
82	243.5	229.0	228.5	239.5	250.0	243.0
83	234.5	221.0	223.0	227.0	242.0	231.5
84	232.0	215.0	213.5	222.0	230.5	229.5
85	245.5	239.5	236.0	248.5	254.5	254.5
86	252.0	245.5	242.0	246.5	255.0	252.0
87	263.5	250.0	253.0	258.5	263.0	259.5
88	266.0	252.5	254.5	261.5	270.5	258.5
89	275.5	277.0	279.0	279.5	280.5	282.0
90	266.0	267.0	266.5	269.0	259.0	267.0

Table B (continued)

	Hours					
	336	504	672	840	1008	1176
1	101.0	104.5	95.0	95.0	91.0	85.0
2	116.5	117.5	111.0	109.0	107.5	98.5
3	119.0	121.0	112.0	111.5	111.0	104.0
4	120.5	119.0	114.0	117.0	112.0	106.5
5	122.5	121.5	115.0	111.5	111.5	107.5
6	126.5	116.5	114.5	112.5	109.5	103.0
7	121.5	121.5	117.0	113.5	115.0	107.5
8	129.0	126.5	119.0	118.5	115.0	108.0
9	124.0	122.5	112.0	114.0	110.5	102.0
10	139.0	135.0	128.5	125.0	117.5	111.0
11	144.0	141.0	129.0	129.5	121.5	109.0
12	140.0	134.5	123.0	121.0	113.0	103.5
13	157.0	150.5	141.5	139.0	132.0	125.5
14	148.0	141.5	131.5	133.5	120.0	116.5
15	151.5	146.0	130.0	129.0	118.5	113.0
16	155.0	150.5	135.5	134.0	123.5	116.5
17	145.0	140.5	124.5	125.5	114.0	105.0
18	152.5	146.5	136.0	131.0	126.0	118.0
19	205.0	195.5	184.0	179.0	192.0	184.0
20	196.5	192.5	173.0	175.0	169.0	166.5
21	130.0	133.0	124.0	122.5	125.5	119.5
22	197.5	197.0	171.5	183.5	188.0	175.5
23	205.0	207.5	186.5	190.0	201.5	190.0
24	195.0	207.5	189.0	189.0	198.0	177.5
25	200.0	200.5	190.0	184.0	186.5	176.0
26	203.0	201.5	192.0	182.5	184.5	175.5
27	191.5	200.5	196.0	183.0	184.0	184.0
28	103.5	94.5	103.0	96.0	114.0	103.5
29	110.0	109.0	111.0	98.5	114.0	103.5
30	126.0	106.5	106.5	99.0	118.0	85.5
31	108.0	102.5	103.5	91.0	94.5	91.0
32	133.5	133.0	135.0	124.0	146.5	144.0
33	136.5	117.5	126.5	123.0	132.5	126.5
34	116.5	114.0	116.0	106.0	120.0	106.0
35	121.5	106.5	110.0	105.5	121.5	111.0
36	120.0	107.5	110.0	106.5	124.0	118.5
37	84.5	84.0	83.5	77.5	88.5	87.0
38	85.5	81.5	79.0	73.0	83.5	78.5
39	85.5	87.5	81.0	77.0	85.0	79.5
40	88.5	83.5	80.5	71.0	83.5	83.5
41	81.0	82.5	77.5	76.5	81.5	82.0
42	89.5	85.0	77.5	76.0	80.0	74.5
43	77.5	79.5	74.0	71.0	76.5	77.5
44	89.0	86.5	75.0	78.0	82.5	80.0
45	92.0	83.0	80.5	76.0	89.5	85.5
46	834.0	715.5	586.0	838.5	804.5	781.0
47	598.5	650.5	422.5	1172.0	1019.5	982.5
48	696.5	734.0	491.0	1034.0	872.0	874.0
49	559.0	566.5	362.5	812.5	702.5	693.0

Table B (continued)

	Hours					
	336	504	672	840	1008	1176
50	615.0	608.0	473.0	730.0	607.5	578.0
51	593.5	578.5	443.0	797.0	649.5	650.0
52	568.0	561.5	425.0	833.5	736.5	701.5
53	471.5	465.0	344.5	823.5	705.0	691.5
54	551.5	530.5	414.5	912.0	766.5	756.0
55	116.0	115.0	107.0	108.5	110.0	121.0
56	106.5	108.0	102.0	100.0	110.5	111.0
57	127.5	117.5	118.0	113.0	115.0	121.0
58	111.5	110.5	104.0	103.0	108.0	112.5
59	109.0	107.5	102.0	105.5	108.0	110.5
60	121.0	118.0	111.5	117.5	118.0	117.5
61	119.0	115.5	110.5	111.5	110.5	116.5
62	109.0	111.5	105.0	105.5	107.0	108.0
63	124.0	117.5	110.5	112.5	110.5	114.0
64	431.0	414.0	372.5	380.0	358.5	360.5
65	406.5	390.0	360.0	350.5	331.5	342.0
66	424.5	415.0	381.5	368.5	353.5	352.0
67	420.0	400.0	369.5	365.0	354.0	335.0
68	425.0	423.0	381.5	382.0	358.0	346.0
69	438.5	435.5	376.5	360.0	351.5	349.0
70	510.0	436.0	379.5	374.0	365.0	339.0
71	455.5	410.0	369.0	359.0	350.0	331.5
72	449.0	402.5	372.0	355.0	351.0	338.0
73	386.0	298.5	241.0	330.0	319.5	264.5
74	317.0	230.5	181.0	422.5	406.5	344.0
75	294.5	196.5	161.5	331.5	313.0	256.5
76	366.5	281.5	234.5	436.0	392.5	366.5
77	369.0	258.0	221.5	390.0	378.5	333.0
78	312.5	224.5	184.5	377.5	358.0	322.5
79	318.5	233.5	192.0	415.0	395.0	362.5
80	241.5	147.5	105.0	389.0	376.0	328.0
81	239.5	160.0	121.5	381.5	360.0	323.5
82	258.0	236.5	227.0	219.5	231.0	229.5
83	242.0	228.0	219.5	214.5	215.5	222.0
84	240.5	222.0	210.0	200.5	211.0	213.0
85	257.0	243.0	228.5	225.5	234.5	231.0
86	256.0	239.5	229.5	222.0	231.0	232.0
87	271.5	247.0	231.0	226.5	236.0	232.0
88	267.5	254.5	223.5	220.5	233.0	237.0
89	285.0	264.5	244.0	233.5	243.0	242.0
90	277.0	253.5	227.0	222.0	223.5	228.5

Table B (continued)

	1344	1512	1680	1848	2016
1	73.5	78.5	65.5	69.5	57.5
2	95.0	98.5	86.5	91.0	84.0
3	102.0	102.0	93.5	101.5	84.5
4	101.0	100.0	88.5	96.0	85.0
5	100.0	102.0	92.5	96.0	87.5
6	105.0	97.5	94.5	90.5	85.0
7	105.0	100.0	93.5	94.5	78.5
8	100.0	103.0	87.5	88.0	83.5
9	97.0	96.0	85.0	84.0	76.0
10	103.5	105.0	87.5	89.5	80.0
11	105.0	102.0	86.0	93.5	75.0
12	91.0	94.0	77.0	82.0	58.5
13	111.5	112.5	96.0	94.5	83.0
14	101.0	103.5	84.5	84.5	66.0
15	96.0	98.5	83.5	83.5	70.5
16	105.0	105.0	90.0	91.5	76.0
17	97.5	96.0	84.0	77.5	64.5
18	111.0	106.0	89.0	91.0	71.0
19	179.5	189.0	165.0	177.0	161.0
20	159.0	159.0	147.5	150.0	137.0
21	124.0	123.0	115.5	119.0	104.5
22	179.0	172.5	164.0	176.5	153.5
23	194.0	191.5	180.5	188.5	186.5
24	185.5	192.0	191.0	192.0	177.5
25	183.0	182.5	185.5	185.0	177.5
26	175.5	167.0	166.5	166.5	149.0
27	185.0	181.5	183.5	184.0	163.0
28	116.0	128.0	109.0	110.0	104.5
29	114.0	128.5	120.0	115.5	121.0
30	99.5	106.0	100.0	104.0	94.0
31	105.5	100.5	92.5	80.0	88.0
32	148.0	156.5	136.5	137.0	131.5
33	141.5	141.0	130.0	143.5	129.0
34	124.0	120.5	104.0	107.5	108.0
35	129.0	130.0	123.0	127.5	109.0
36	124.0	117.5	114.0	117.0	119.5
37	85.5	92.0	85.0	78.5	78.5
38	83.0	84.0	78.5	78.5	77.0
39	82.0	91.0	86.0	83.0	82.0
40	83.5	83.0	87.0	86.0	80.0
41	85.0	86.0	81.0	81.0	76.5
42	81.5	82.0	77.0	83.0	80.0
43	77.5	75.5	73.5	74.5	72.0
44	86.0	87.5	78.5	83.5	81.0
45	86.0	88.5	81.0	79.5	85.0
46	752.5	531.5	479.0	425.0	307.0
47	902.0	754.0	669.5	651.0	497.0
48	755.5	682.5	648.0	624.5	489.0
49	533.5	396.0	413.0	446.5	284.5

Table B (continued)

	1344	1512	1680	1848	2016
50	445.5	368.5	307.0	370.0	181.5
51	523.5	418.5	368.5	395.0	234.5
52	548.5	463.0	430.5	443.5	307.5
53	514.0	458.5	416.0	434.0	274.5
54	609.0	506.5	470.0	483.0	248.0
55	110.0	107.5	117.5	118.0	109.5
56	106.5	109.0	116.5	113.0	105.5
57	116.0	119.0	119.0	116.5	115.5
58	109.5	114.0	110.0	113.0	102.0
59	106.5	109.5	106.5	112.0	105.0
60	114.5	120.0	118.0	116.5	113.0
61	114.0	115.0	115.0	119.5	110.0
62	107.0	108.0	107.5	113.0	101.0
63	111.5	113.0	118.0	115.5	105.5
64	354.5	349.5	342.0	351.5	309.0
65	335.0	325.5	312.5	342.5	283.5
66	339.5	340.0	321.0	328.5	298.0
67	333.0	320.0	321.5	336.5	291.0
68	332.0	321.5	319.0	337.5	288.0
69	323.0	317.0	314.5	318.5	288.0
70	361.5	357.5	340.0	342.5	317.5
71	335.5	343.0	325.5	311.5	317.0
72	337.0	355.5	318.5	315.0	313.5
73	270.0	238.5	175.5	121.5	97.5
74	349.5	297.0	268.5	225.5	162.5
75	282.0	199.0	195.5	151.5	100.0
76	376.0	321.0	305.5	241.0	200.5
77	321.5	273.0	257.0	191.5	152.5
78	319.0	274.0	241.0	194.0	150.0
79	361.0	314.5	282.5	213.0	189.5
80	332.5	263.0	244.5	172.5	147.5
81	312.0	255.5	248.0	202.5	149.0
82	227.0	231.0	228.5	224.5	214.5
83	221.0	219.5	221.0	204.5	209.0
84	211.5	211.5	208.5	200.5	202.5
85	229.0	231.0	223.0	215.0	217.0
86	228.5	228.5	232.5	222.0	217.0
87	230.5	236.0	229.0	219.5	213.0
88	236.0	236.0	229.0	222.0	216.0
89	244.5	242.0	234.5	223.0	218.5
90	228.0	222.5	222.0	214.0	202.0

Table B (continued)