Bending Properties of Co–Ni–Cr–Mo Alloy Wire for Orthodontic Application

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In order to evaluate the clinical performance of the Co–Ni–Cr–Mo alloy wire in orthodontic application, bending properties of Co–Ni–Cr–Mo alloy wire was investigated in comparison with the conventional stainless steel wire and Co–Cr alloy wire. Three-point bending test was carried out for the wires with varied drawing rate up to 3.0 mm in deflection and then unloaded with 0.2 mm/s of loading/unloading speed. Bending yield load of the Co–Ni–Cr–Mo alloy wire increased and the residual deflection decreased with increasing wiredrawing rate. The best processing condition was decided as the wiredrawing rate of 78% with the age-hardening treatment. The bending elastic modulus of the Co–Ni–Cr–Mo alloy wire was higher than those of stainless steel wire and Co-Cr alloy wire. The bending strength of the Co–Ni–Cr–Mo alloy wire was lower than that of stainless steel wire and was comparable to that of Co–Cr alloy wire.

Keywords: cobalt nickel chromium molybdenum alloy, orthodontic wire, dental application, three-point bending test, drawing rate, heat treatment, bending yield, bending strength, bending elastic modulus

1. Introduction

Many types of metallic materials are used for many dental applications in restorative dentistry, prosthetics, orthodontics and oral surgery, such as gold alloys and Co–Cr alloys for dental castings, titanium and titanium alloys for dental implants and bone plates, etc. Among the varieties of clinical purposes of metallic materials in dentistry, orthodontic wires are used to produce the force to move mal-aligned teeth to their proper positions. Although the main orthodontic force derives from elastic or super-elastic recovery in bending, torsional moment also works for correcting tooth rotation by means of rectangular wires. These orthodontic wires are connected to the slots of orthodontic brackets adhered to each tooth.

The main alloys of orthodontic wires are stainless steel, Co–Cr alloy, Ti–Ni alloy and Ti–Mo alloy. The wire selection by orthodontists is based on the clinical purpose and treatment method. Recently, super-elastic Ti–Ni alloy wires are widely used for the early dynamic stage, where relatively large tooth movement is required. Ti–Ni alloy wires are capable of exhibiting a super-elasticity which provides light, continuous force for physiological and efficient tooth movement.1,2

On the other hand, high elastic modulus is required in the finishing stage to control the final slight movement of teeth. Hard stainless steel wires and Co–Cr alloy wires are normally used for this purpose. However, the former are easily broken and have rough surface, while the latter need to be heat-treated after bending in clinics to be hardened and consequently discolor.

It was reported that a Co–Ni–Cr–Mo alloy, developed as a new spring material, possessed superior mechanical properties and high corrosion resistance.3) The purpose of this study is to investigate the bending properties of this Co–Ni–Cr–Mo alloy wire in comparison with conventional orthodontic wires with high elastic modulus in order to evaluate the clinical performance of the Co–Ni–Cr–Mo alloy wire in orthodontic application.

2. Experimental Procedures

Co–Ni–Cr–Mo alloy wires (SPRON 510, SII Micro Parts Ltd., Japan), 0.402 mm in diameter, were used in this study. The chemical composition of the alloy was; Ni: 30, Cr: 21, Mo: 10, Fe: 2, Nb: 1.5, Ti: 0.8, Mn: <0.5, Bal: Co (mass%). Three wiredrawing rates were applied to process the wires: 70, 78 or 86%, then straightened. Age-hardening treatment at 873 K for 7.2 ks in vacuum was performed. For the references, a hard stainless steel wire (Australian Wire, Special plus, TP Orthodontics Inc., USA) and a Co–Cr alloy wire (Red Elgiloy, Resilient, RMO, USA) were used, of which the specimen diameters were 0.412 and 0.400 mm, respectively. Heat treatment was performed to harden the Co–Cr wire in a clinical condition, since this treatment is usually performed to harden this wire after forming.

To investigate the bending property of the wires a three-point bending test was carried out, which simulates the clinical application of the wire on teeth.2,4,5) Figure 1 shows a

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Fig. 1 Schematic drawing of the three-point bending test device, viewed from above.
schematic drawing of the device. The center pole was combined with a load cell to measure the load on the wire. The two side poles were mounted on a movable stage connected with a displacement transducer to measure the deflection of the wire. Neither bracket nor ligature wire was used to avoid the influence of ligature condition. The maximum deflection was set at 3.0 mm, then the load on the wire was removed. The loading and unloading speed was approximately 0.2 mm/s.

To compare the bending property of the wires, one-way factorial analysis of variance was used for the detection of the differences among conditions. Tukey-Kramer test was performed as the post hoc test for the detection of the differences between conditions. Statistical significance was set at $p < 0.01$.

3. Results and Discussion

To evaluate the clinical performance of the new Co–Ni–Cr–Mo alloy wire, a bending test simulated the clinical condition was selected, because the main orthodontic force for tooth movement was derived from the elastic deformation in bending of orthodontic wires. Typical load-deflection diagrams of the Co–Ni–Cr–Mo alloy wires with different wiredrawing rate before and after age-hardening treatment at 873 K for 7.2 ks are shown in Fig. 2. Within each specimen group with or without age-hardening treatment, yield load increased and residual deflection decreased with increasing wiredrawing rate. With respect to the influence of age-hardening, elastic modulus and yield load increased markedly and residual deflection decreased in addition to the above-mentioned changes. Since the age-hardened wire with wiredrawing rate of 86% showed rather brittle property in a trial bending test to form clinical loops, the age-hardened wire with wiredrawing rate of 78% was selected to be compared with the referential wires.

Figure 3 shows typical load-deflection diagrams of the age-hardened Co–Ni–Cr–Mo alloy wire with wiredrawing rate of 78%, the hard stainless steel wire, and the heat-treated Co–Cr alloy wire. To eliminate the influence of the difference in specimen diameter, three parameters were taken from these diagrams, bending elastic modulus, bending strength, and residual deflection to compare the bending properties of the alloy wires.

The bending elastic moduli of the age-hardened Co–Ni–Cr–Mo alloy wire with wiredrawing rate of 78%, the hard stainless steel wire, and the heat-treated Co–Cr alloy wire are shown in Fig. 4. The bending elastic modulus is a parameter...
exhibiting the functional force level in orthodontic treatment, and high value in this modulus is effective at the final treatment stage where the slight correction of teeth alignment is required. The value of the age-hardened Co–Ni–Cr–Mo alloy wire was significantly higher than those of the hard stainless steel wire and the heat-treated Co–Cr alloy wire. Therefore, the functional property of the new Co–Ni–Cr–Mo alloy wire was evaluated to be superior to the wires currently used.

Bending strength of the wires was calculated with the maximum bending load within the deflection range up to 3.0 mm. The bending strength values of the age-hardened Co–Ni–Cr–Mo alloy wire with wiredrawing rate of 78%, the hard stainless steel wire, and the heat-treated Co–Cr alloy wire are shown in Fig. 5. Since the orthodontic force is delivered in the elastic deformation range of the wires, large stress exceeding the elastic limit is applied only in the forming process of the wires to fit the individual dental arch. Therefore, this value is a parameter to show the required force in shaping as well as the maximum orthodontic force of the wires. Although the bending strength value of the Co–Ni–Cr–Mo alloy wire was the lowest among the three alloy wires, it was significantly lower only than that of the hard stainless steel wire. This means a relatively low elastic stress range and easy shaping by orthodontists of the Co–Ni–Cr–Mo alloy.

Figure 6 shows the residual deflection, after being unloaded from 3.0% deflection, of the age-hardened Co–Ni–Cr–Mo alloy wire with wiredrawing rate of 78%, the hard stainless steel wire, and the heat-treated Co–Cr alloy. This value shows the amount of permanent deformation after fixed condition of deformation and means the workability of the wires. The residual deflection value of the age-hardened Co–Ni–Cr–Mo alloy wire was significantly higher than those of the other two, the hard stainless steel wire and the heat-treated Co–Cr alloy wire, while the value of the hard stainless steel wire was also significantly higher than that of the heat-treated Co–Cr alloy wire. Consequently, it was evaluated that the workability of the new Co–Ni–Cr–Mo alloy wire was superior to the other wires.

Since orthodontic wires are used in the oral cavity of patients for months, corrosion resistance is a very important factor for the material. It was reported that this Co–Ni–Cr–Mo alloy showed much better corrosion resistance than the stainless steels of SUS 304 and 316L in chemically accelerated conditions. Since SUS 304 and 316L are major materials for orthodontic wires and surgical implants, respectively, the corrosion resistance of the Co–Ni–Cr–Mo alloy is thought to be sufficient for the dental application.

Co–Cr alloys have been used as major metallic materials in dentistry and surgery for a long time. The composition of the alloy used in this study is rather similar to the one of an alloy for surgical implant, Co–Ni–Cr–Mo alloy. The composition of the Co–Ni–Cr–Mo alloy used in this study is different from the one in the standards in low nickel content and niobium addition. The Co–Cr alloy wire used as a referential material in this study also included additional elements. The approximate basic compositions is reported to be Co–20Cr–16Fe–15Ni–7Mo.

Among the dental and surgical alloys with similar constituent elements, the superior bending properties of the Co–Ni–Cr–Mo alloy used in this study is caused by the alloy composition as well as the selection of the conditions for work hardening and aging treatment. In addition, the Co–Ni–Cr–Mo alloy wire showed clinically better surface condition than the references, smooth surface to follow the tooth movement for better clinical performance without being discolored by additional heat treatment.

4. Conclusions

The bending properties of the new Co–Ni–Cr–Mo alloy wires were investigated in comparison with conventional
hard stainless steel wires and the heat-treated Co–Cr alloy wires. It was evaluated that the Co–Ni–Cr–Mo alloy wire showed superior bending properties as an orthodontic material as follows:

1) Yield load of the Co–Ni–Cr–Mo alloy wire increased and the residual deflection decreased with increasing wiredrawing rate and age-hardening treatment at 873 K for 7.2 ks. The best processing condition was decided as the wiredrawing rate of 78% with the age-hardening treatment.

2) The bending elastic modulus of the Co–Ni–Cr–Mo alloy wire was higher than those of the hard stainless steel wire and the heat-treated Co–Cr alloy wire.

3) The bending strength of the Co–Ni–Cr–Mo alloy wire was lower than that of the hard stainless steel wire and was comparable to that of the heat-treated Co–Cr alloy wire.

4) The residual deflection of the Co–Ni–Cr–Mo alloy wire was higher than those of the hard stainless steel wire and the heat-treated Co–Cr alloy wire.

It is concluded that the Co–Ni–Cr–Mo alloy wire is evaluated to exhibit excellent clinical performance and handling for the final stage of orthodontic treatment.

REFERENCES