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Influence of normal forces on the frictional behavior in tribological systems made of different bracket types and wire dimensions

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The aim of the present work was measuring the effect of varying normal forces on frictional forces applied to different bracket types in combination with archwires made of NiTi and stainless steel of variable cross section. The measurements were carried out in artificial saliva. Three-way ANOVA and Bonferroni *post-hoc* tests (α=0.05) were applied. Except for one subgroup the combination of normal force, bracket system and wire dimension had significant effect on friction (p<0.001) as friction increased with increasing normal forces. Only moderately tied ligatures or passive self-ligating brackets generate low friction forces. There was a statistically significant order (0.016"×0.022"<0.018"×0.025"<0.019"×0.025") for stainless steel wire material. Finite element modeling simulation showed the increasing effect of active clip force on friction especially for 0.025" wire profiles. If compared to NiTi wires, stainless steel archwires delivered higher friction. Combinations between wire-type and ligation should be chosen carefully for the intended treatment step.

Keywords: Friction, Normal force, Orthodontic ligature, Artificial saliva

INTRODUCTION

The interaction of archwires with surfaces of bracket slots and ligatures results in frictional forces. Strictly speaking, the reason for the occurrence of frictional forces are the applied normal forces which press the archwire against the bottom of the slot, possibly combined with other less well-defined forces resulting from binding, notching or from the different ligatures used[1]. When the clinical forces applied overcome friction between bracket and wire, tooth movement takes place[2]. The contact between bracket and wire is a tribological system; a system of two contacting bodies and an intermediary fluid relevant to the friction between the two bodies[1].

Friction may reduce the total functional force necessary for tooth movement by about 50%[3-6]. Therefore, high levels of friction are undesirable; even worse, they can cause loss of anchorage[7-10]. Moreover, higher friction may lead to pain and tissue damage or failure of gap closure[11-13]. Consequently, knowledge and control of the parameters influencing the magnitude of frictional forces during tooth movement is important. Physically, frictional force F_f is proportional to normal force F_N, linked by the frictional coefficient μ. The frictional equation shows the following simple relation

$$F_f=\mu \cdot F_N$$

When describing frictional behavior of orthodontic appliances, it may be necessary to consider further forces. This will extend the equation to:

$$F_R=\mu_1 \cdot F_N+\mu_2 \cdot F_{Lig.}+F_{B/N},$$

where $F_{B/N}$ describes the contribution of binding and notching to the total frictional force, while μ_1 and μ_2 describe the frictional coefficients of the two involved tribological systems. The normal force F_N acts perpendicularly to the contact surfaces. In orthodontics, the ligation's force $F_{Lig.}$ can be considered as the main contribution to the normal force. In most studies, high normal forces produced by ligation will increase frictional forces[14-16]. To understand frictional mechanics, normal force is a very important aspect to consider. Different methods of ligation, such as conventional steel ligations, elastomeric ligations on classical twin brackets or self-ligating brackets result in varying normal forces and their corresponding frictional forces. Self-ligating brackets offer lower friction and high treatment effectivity is advertized[2,4,12,17]. Additionally, the self-ligating concept makes the use of steel ligatures obsolete due to their inherent latch mechanism, which can be opened and closed to secure the wire. In this way, self-ligating brackets facilitate the archwire removal and eliminate the uncertainties related to ligation forces caused by ligature wires applied by different clinicians[18].

Depending on the ligature mechanism of self-ligating brackets, two main groups can be distinguished: active and passive self-ligating brackets. Active self-ligating bracket systems have a slot-closure system, similar to a clip that closes the slot from the top (labial-buccal). The active spring clip is designed to exert an additional force on the archwire when the suitable wire dimension is used, whilst the additional force is being stored in the deflection of the clip. In contrast, passive self-ligating bracket systems have a latch that acts like a door turning the slot into a tube. This kind of slide does not lead to active pressure on the archwire.

Thus, it is worth considering the difference between active and passive self-ligating brackets in detail. The

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aforementioned studies evaluated the performance of passive slides and active clips of self-ligating mechanism from the aspects of alignment, friction and ease of use. In comparison, the initial alignment can be more complete using an active clip for a particular wire size.

When combining a stiff and large sized rectangular wire with active self-ligating brackets, the active clip will increase friction during tooth movement[1]. In contrast, the increased play between the wire and the passive slot will generate lower forces and may release the frictional force caused by adjacent tilted teeth[18]. Still, it is controversial whether passive self-ligating brackets reduce treatment time or rather alignment efficiency. While some argue there is no reduction in treatment time[19], there are studies that found active self-ligating brackets to be more efficient for alignment[20]. So far there is no evidence to favor self-ligating fixed appliances over conventional ones in the following clinically significant variables[21,22]. These may include but are not limited to subjective pain, bond failure, arch dimension changes, time needed for space closure, periodontal outcomes, or root resorption.

In addition, the wire material is one of the most important factors influencing the friction force. However, the influence of the wire material is controversially discussed in the literature[6,8,13,23-33]. The discrepancy in performance may result from the wire materials themselves as well as from their manufacturing process[29].

One has to consider all of the above-mentioned parameters during treatment planning. As different treatment steps or objectives have different requirements, the role of friction must be viewed differently for each treatment objective or treatment phase. During alignment, frictional forces of unknown magnitude and direction will cause unwanted side effects and lead to a less predictable outcome. In contrast to that, torque control needs minimal play of the archwire in the slot, which may be related to additional frictional forces.

Purpose of the present study was to investigate the effect of different normal forces on the total frictional forces occurring in the tribological systems consisting of conventional, passive and active self-ligating brackets made of ceramics as well as from stainless steel in combination with wires made of NiTi and stainless steel. For each of the mentioned wire materials and bracket categories, samples were used from established market product. This choice of samples reflects the state of the art in orthodontic industry and is not intended to be a product test.

The paper also analyzes the effect of contact forces between latches of active self-ligating brackets and archwire surfaces by means of calculations based on finite element modeling (FEM).

MATERIALS AND METHODS

The present study was conducted on 24 ceramic active self-ligating brackets (In-Ovation® C, Dentsply Sirona®, Islandia, NY, USA), 24 metal active self-ligating brackets (BioQuick®, Forestadent®, Pforzheim, Germany), 24 metal passive self-ligating brackets (Damon™ 3MX, Ormco™, Glendora, CA, USA) and 24 conventional twin brackets (Mini Sprint®, Forestadent®) with a 0.022" slot size (Table 1). The work was intentionally carried out using standard market products as friction is considered

Table 1 Brackets and wires used in this study

Product	Manufacturer	Material	Dimension	Ligation	Abbreviations	N
Mini Sprint®	Forestadent®	X8CrMnMoN18-18-2 (correspon-ding AISI classification n.a.)	0.022"	Ligature wire	M-TB (Metal twin bracket)	24
In-Ovation® C	Dentsply Sirona®	Polycrystalline Al_2O_3	0.022"	Active self-ligating	A-C-SL (Active Ceramic SL)	24
BioQuick®	Forestadent®	X8CrMnMoN18-18-2 (correspon-ding AISI classification n.a.)	0.022"	Active self-ligating	A-M-SL (Active Metal SL)	24
Damon™ 3MX	Ormco™	X5CrNiCuNb17-4-4 (corresponds to AISI 630)	0.022"	Passive self-ligating	P-M-SL (Passive Metal SL)	24
Steel arch wires	Forestadent®	X10CrNi18-8 (corresponds to AISI 302)	0.016"×0.022", 0.018"×0.025", 0.019"×0.025"	—	ss	48 (16 each)
BioTorque® Arches	Forestadent®	Superelastic NiTi alloy, ca. 50.8 at% Ni	0.016"×0.022", 0.018"×0.025", 0.019"×0.025"	—	NiTi	48 (16 each)

being a system variable rather than a material property. Therefore, one has to define a representative tribo system consisting of the frictional bodies (bracket and wire) and the intermediate fluid (artificial saliva). The chosen brackets and wires are typical for their material class.

Prescriptions of the brackets with torque and angulation were compensated in the testing machine by the use of multiple linear, rotation and goniometric stages (see Figs. 1 and 2). In total, 48 stainless steel and 48 NiTi orthodontic rectangular archwires exhibiting dimensions of 0.016"×0.022", 0.018"×0.025", 0.019"×0.025" (Table 1) were included. All tests were performed using artificial saliva at (36±1)°C to simulate the oral environment during orthodontic treatment. Temperature control was achieved by a proportional-integral-derivative (PID) thermostat REX-C100 (RKC Instrument, Saitama, Japan). Artificial saliva was applied by means of a syringe and a flexible tube onto the bracket-wire tribological system with a dripping rate of about 3 mL/min. The measurements were conducted using different magnitudes of normal force F_N of 1.0 N, 1.5 N, 2.0 N, 2.5 N. All archwires and brackets used in this study were cleaned with 95% ethanol and air-dried prior to testing in order to remove organic residues.

The measurement program was developed by the use

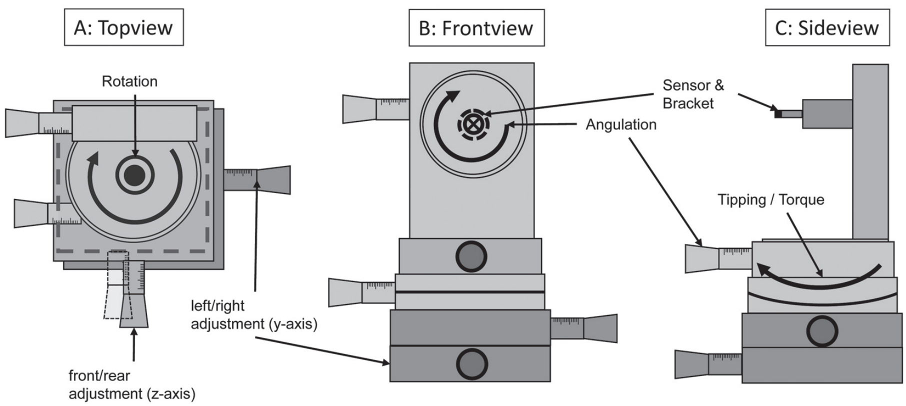

Fig. 1 Schematic experimental setup showing the adjustment rig used to level out the prescriptions and possible side effects like binding or notching.
A: Topview of the rig. Front-rear offset (z-axis), left-right offset (y-axis) and rotation can be adjusted. B: Frontview, showing the possibility to adjust angulation. C: Sideview, used to level out torque. All adjustments are performed by hand and observed with magnifiers (10×).

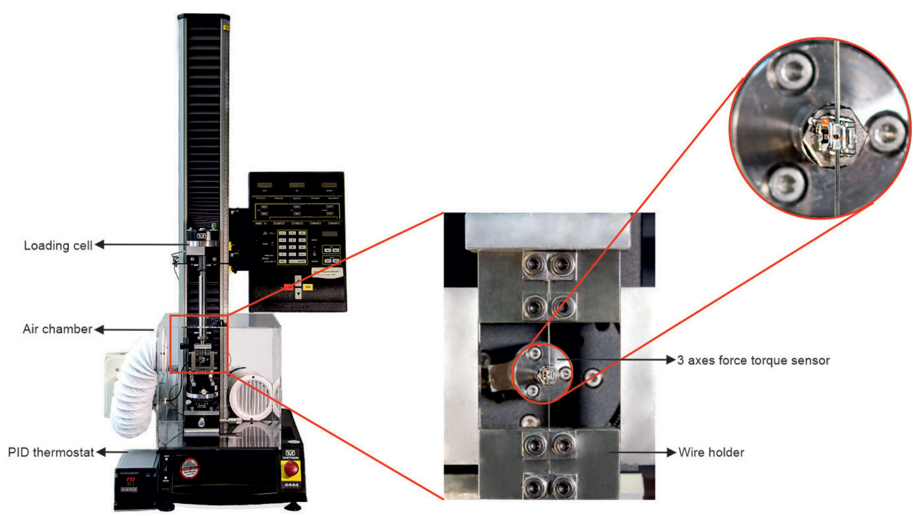

Fig. 2 Experimental setup used in this study: The bracket-wire combination was combined with a sensor and a load cell.
All measurements were conducted at (36±1)ºC in a temperature-controlled chamber (REX-C100 PID thermostat).

of Labview 2012® (National Instruments, Austin, Texas, USA). To pull the wire through the sample brackets a universal testing machine (Instron® 4444, Instron®, Norwood, MA, USA) was used in this study. The bracket samples were mounted to a six axes force/torque (F/T) sensor (Nano 17 SI-25-0.25, ATI, Industrial Automation, Apex, NC, USA). The archwire sections were mounted into the test setup by means of two clamps. Before applying any external force, alignment of bracket and wire was verified by conducting a zero-force run where the sensor was used to detect any possible misalignment, which then was leveled out by readjustment of the system. Thus, unintentional frictional forces between slot walls and arch wire were avoided. Afterwards, the wire was locked into the bracket by applying the ligature, which was either done manually or by latch mechanism. The sensor exhibits a measurement range of ±25 N and a resolution of 0.007 N. The sampling rate was set to 10 Hz. The sensor is subjected to a regular calibration procedure.

During the measurement, the wire was moved over a distance of 3 mm at the speed of 10 mm/min through the slot of the bracket using the actuation from the universal testing machine[1]. Frictional force values were recorded by the force/torque sensor. The initial peak value of the frictional force was considered as static frictional force. The calculation of the kinetic friction was performed on the average force delivered during the first 1 mm after static frictional force peak (Fig. 3).

In addition, the bracket slot dimensions were measured by means of an AmScope SM-1TNZ-144A stereo microscope using a calibrated 10MPixel camera (MU1000) and the original device software in Rev. 4.8 (2019) (United Scope, Irvine, CA, USA). The wire dimensions (height×width) were measured with a Mahr 40EX (Mahr, Providence, RI, USA) calibrated micrometer gauge with a resolution of 0.001 mm.

Scanning electron microscope (SEM) techniques were used to analyze the surface conditions of the different brackets as well as arch wires (SEM, ZEISS Supra 55vp, Zeiss, Oberkochen, Germany). All samples were sputter coated with a thin layer of gold-palladium alloy (SC7620, Polaron, Quorum Technologies, Kent, UK) after ultrasonic cleaning in Ethanol. The acceleration voltage was set to 10 kV and secondary electrons were used for image generation. Special focus was set to surface quality inside the bracket slots as well as identification of burrs at the edges of the slot which possibly could increase friction.

The software ANSYS 2020 R2 (Ansys, Canonsburg, PA, USA) was used to calculate the amount of force generated by the active bracket latch as a function of archwire dimensions. The latch dimensions were measured using the before mentioned light microscope and was reverse-engineered using Autodesk Inventor 2020 (Autodesk, San Rafael, CA, USA) and transferred to Ansys. The meshed model consisted of SOLID 187 elements using an element size of 0.1 mm. The built-in material parameters for stainless steel 316L were used with the Young's modulus E(36°C)≈1.94 GPa and the Poisson's ratio of ν(36°C)≈0.25. A temperature of 36°C was simulated to match the *in vitro* experiments. In order to minimize the influence of the experimental error, all samples were tested in random order. Each sample was measured 5 times. Normal distribution of the data was tested using the Kolmogorov-Smirnov test. A three-way ANOVA was used to compare and indicate significant differences between the impact of normal force, bracket system and archwire dimension on the frictional values. *Post-hoc* tests were calculated if the result was significant. Bonferroni correction was applied to correct for multiple testing. Calculations were carried out using SPSS 25 (IBM, Armonk, NY, USA), with a significance level of α=0.05.

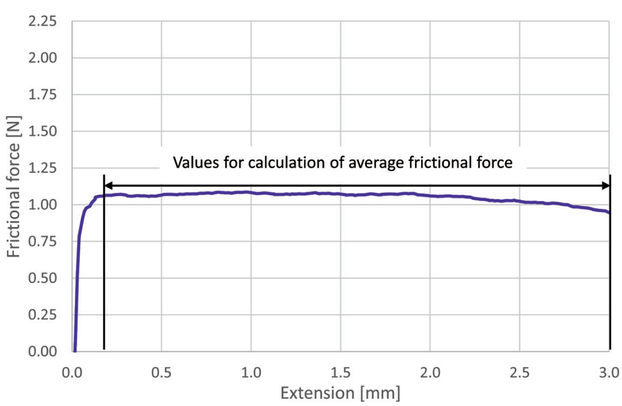

Fig. 3 Typical frictional force-deflection curve.

RESULTS

For both wire materials, stainless steel as well as nickel titanium, almost all of the three variables (normal force, bracket system and wire dimension) and their interactions had significant effects on frictional force measured according to three-way ANOVA (p<0.001). There was one exception when using archwires made of nickel titanium, where the interaction of normal force and wire dimension did not have a significant effect on friction (p=0.217).

A typical frictional force-deflection curve is shown in Fig. 3. The initial peak is attributed to the so-called static friction, which is an amount of mechanical energy needed to overcome the static adhesive forces on a microscopic level between two surfaces in contact with each other.

Overall, there was a positive relation between normal force and resulting frictional force (Fig. 4). Even a relatively small increase of the normal force resulted in a relatively high frictional force. For passive metal self-ligating brackets in combination with a NiTi wire dimension of 0.016"×0.022", the frictional force values increased more than threefold from 0.199 N (F_N=1.0 N) to 0.616 N (F_N=2.5 N). The same was the case using 0.016"×0.022" stainless steel wires: the frictional force increased from 0.296 N (F_N=1.0 N) to 0.715N (F_N=2.5

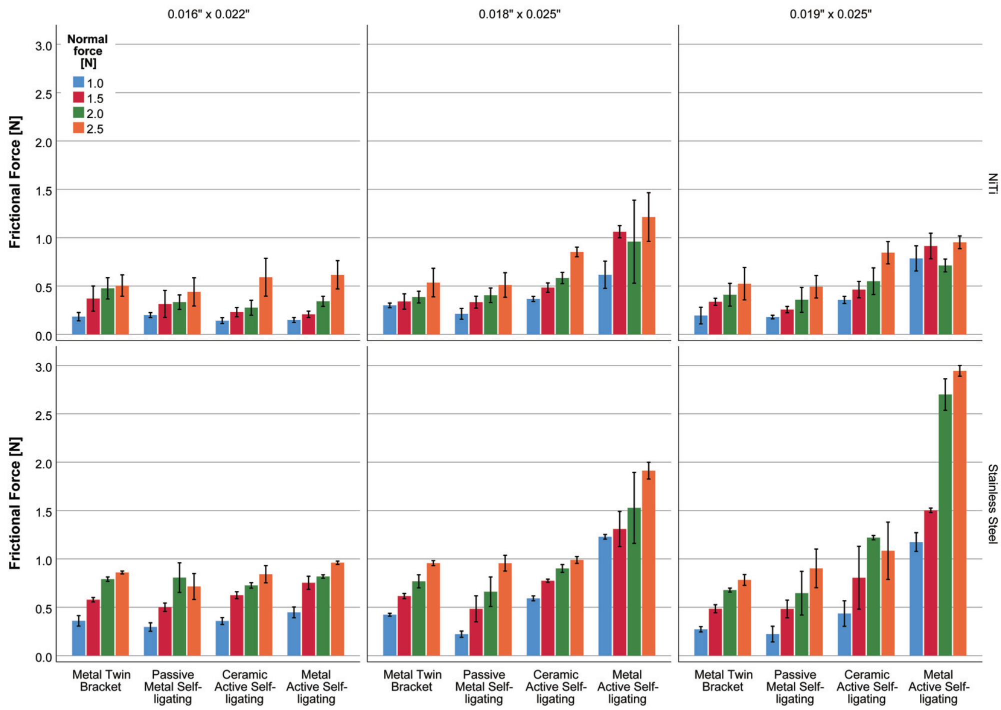

Fig. 4 Mean normal force±standard deviation for all tribological systems tested in this study.

Table 2 Kinetic friction [F] for tribological systems made from different normal forces (F_N), NiTi wire dimensions and bracket systems. The bracket system used are: Damon™ 3MX (Ormco™) passive metal self-ligating (P-M-SL), In-Ovation® C (Dentsply Sirona®), active ceramic self-ligating (A-C-SL), Mini Sprint® (Forestadent®) metal twin brackets (M-TB) and BioQuick® (Forestadent®) active metal self-ligating brackets (A-M-SL)

F_N [N]	Dimension (inch)	Bracket system			
		P-M-SL	A-C-SL	M-TB	A-M-SL
1.0	0.016×0.022	0.199 (0.020)^{AaW}	0.142 (0.025)^{AaW}	0.183 (0.035)^{AaW}	0.148 (0.020)^{AaW}
	0.018×0.025	0.212 (0.035)^{AaW}	0.367 (0.021)^{BbW}	0.300 (0.019)^{ABbW}	0.616 (0.113)^{CbW}
	0.019×0.025	0.179 (0.015)^{AaW}	0.356 (0.030)^{BbWX}	0.195 (0.054)^{AabW}	0.786 (0.105)^{CcW}
1.5	0.016×0.022	0.315 (0.088)^{AbaX}	0.230 (0.039)^{AaWX}	0.370 (0.082)^{BaX}	0.207 (0.027)^{AaW}
	0.018×0.025	0.333 (0.049)^{AaX}	0.484 (0.039)^{BbX}	0.340 (0.064)^{AaW}	1.061 (0.051)^{CcX}
	0.019×0.025	0.256 (0.026)^{AaWX}	0.463 (0.069)^{BbXY}	0.337 (0.030)^{AaX}	0.914 (0.106)^{CbX}
2.0	0.016×0.022	0.333 (0.060)^{AaXY}	0.276 (0.062)^{AaX}	0.476 (0.088)^{BaXY}	0.342 (0.042)^{AaX}
	0.018×0.025	0.404 (0.061)^{AaXY}	0.583 (0.047)^{BbX}	0.386 (0.049)^{AaW}	0.959 (0.269)^{CcX}
	0.019×0.025	0.357 (0.104)^{AaX}	0.550 (0.111)^{BbY}	0.410 (0.074)^{AaX}	0.713 (0.053)^{CbW}
2.5	0.016×0.022	0.439 (0.092)^{AaY}	0.591 (0.158)^{BCaY}	0.505 (0.089)^{AbaY}	0.616 (0.118)^{CaY}
	0.018×0.025	0.511 (0.102)^{AaY}	0.852 (0.031)^{BbY}	0.536 (0.118)^{AaX}	1.213 (0.203)^{CcY}
	0.019×0.025	0.493 (0.094)^{AaY}	0.845 (0.073)^{BbZ}	0.524 (0.135)^{AaY}	0.952 (0.053)^{BbX}

Three-way ANOVA was carried out as described. Upper-case letters A, B, C designate differences between bracket systems for each combination of normal force and wire dimension. Lower-case letters indicate differences between wire dimensions for each combination of bracket system and normal force. Values with differing W, X, Y, Z are statistically different when looking at the effect of normal force for each combination of wire dimension and bracket system.

N). While employing a normal force of 1.0 N produced the lowest, employing a normal force of 2.5 N produced the highest frictional forces (Fig. 4). These were in turn significantly different from each other (Tables 2 and 3). This effect is less clear for normal forces of 1.5 N and 2.0 N that did not always produce a clear ranking nor a statistically different resultant frictional force (Fig. 4, Tables 2 and 3).

Table 3 Kinetic friction [F] for tribological systems made from different normal forces (F_N), stainless steel wire dimensions and bracket systems. The bracket system used are: Damon™ 3MX (Ormco™) passive metal self-ligating (P-M-SL), In-Ovation® C (Dentsply Sirona®), active ceramic self-ligating (A-C-SL), Mini Sprint® (Forestadent®) metal twin brackets (M-TB) and BioQuick® (Forestadent®) active metal self-ligating brackets (A-M-SL).

F_N (N)	Dimension (inch)	Bracket system			
		P-M-SL	A-C-SL	M-TB	A-M-SL
1.0	0.016×0.022	0.296 (0.035)^{AaW}	0.358 (0.029)^{ABaW}	0.359 (0.043)^{ABabW}	0.447 (0.045)^{BaW}
	0.018×0.025	0.222 (0.026)^{AaW}	0.592 (0.020)^{CbW}	0.423 (0.012)^{BbW}	1.228 (0.019)^{DbW}
	0.019×0.025	0.223 (0.065)^{AaW}	0.435 (0.106)^{BbW}	0.272 (0.022)^{AaW}	1.174 (0.078)^{CcW}
1.5	0.016×0.022	0.500 (0.035)^{AaX}	0.625 (0.029)^{BaX}	0.578 (0.019)^{AabX}	0.753 (0.054)^{CaX}
	0.018×0.025	0.484 (0.108)^{AaX}	0.774 (0.013)^{CbX}	0.616 (0.022)^{BbX}	1.309 (0.146)^{DbW}
	0.019×0.025	0.483 (0.073)^{AaX}	0.805 (0.262)^{BbX}	0.484 (0.033)^{AaX}	1.503 (0.018)^{BcX}
2.0	0.016×0.022	0.806 (0.123)^{AbY}	0.726 (0.023)^{AaXY}	0.790 (0.018)^{AaY}	0.818 (0.015)^{AaX}
	0.018×0.025	0.661 (0.123)^{AaY}	0.901 (0.033)^{BbY}	0.768 (0.055)^{AaY}	1.527 (0.296)^{CbX}
	0.019×0.025	0.645 (0.182)^{AaY}	1.220 (0.018)^{BcY}	0.677 (0.017)^{AaY}	2.700 (0.131)^{CcY}
2.5	0.016×0.022	0.715 (0.108)^{AaY}	0.842 (0.72)^{BaY}	0.859 (0.012)^{BabY}	0.960 (0.014)^{CaY}
	0.018×0.025	0.956 (0.066)^{AbZ}	0.989 (0.029)^{AbY}	0.956 (0.020)^{AaZ}	1.913 (0.070)^{BbY}
	0.019×0.025	0.902 (0.161)^{AbZ}	1.084 (0.239)^{CbZ}	0.782 (0.045)^{BbY}	2.945 (0.045)^{DcZ}

Three-way ANOVA was carried out as described. Upper-case letters A, B, C designate differences between bracket systems for each combination of normal force and wire dimension. Lower-case letters indicate differences between wire dimensions for each combination of bracket system and normal force. Values with differing W, X, Y, Z are statistically different when looking at the effect of normal force for each combination of wire dimension and bracket system.

Table 4 Results of dimensional measurements on arch wires (all dimensions in micrometer). The last column gives a feedback about the conformity of the measured data with the ISO 13996 standard for arch wire dimensions

Material		Nominal dimensions		Measured dimensions	Deviation	In relation to industrial norm (DIN EN ISO 13996)
				Mean±SD	nominal *vs.* measured	
		(inch)	(µm)	(µm)	(%)	
stainless steel	Height	0.016	406.4	426.1±6.4	4.6	above specification
	Width	0.022	558.8	580.9±7.3	3.8	above specification
	Height	0.018	457.2	461.8±3.2	1.0	meets specification
	Width	0.025	635.0	639.3±9.4	0.7	meets specification
	Height	0.019	482.6	487.4±7.4	1.0	meets specification
	Width	0.025	635.0	640.3±5.8	0.8	meets specification
NiTi	Height	0.016	406.4	432.6±4.0	6.0	far above specification
	Width	0.022	558.8	583.4±2.1	4.2	far above specification
	Height	0.018	457.2	466.3±5.6	2.0	meets specification
	Width	0.025	635.0	647.4±7.1	1.9	meets specification
	Height	0.019	482.6	491.7±7.8	1.9	above specification
	Width	0.025	635.0	641.1±4.2	1.0	meets specification

Bracket design influenced the resulting frictional force significantly ($p<0.001$). In combination with large sized rectangular archwires made of either stainless steel or NiTi, the passive self-ligating brackets used in this study produced the lowest frictional values: 0.212 N (ss), 0.222 N (NiTi) for 0.018"×0.025" wires and 0.179 N (ss), 0.223 N (NiTi) for 0.019"×0.025" wires (Tables 2 and 3) when a normal force of 1.0 N was employed. With the same normal force, the active metal self-ligating brackets had the highest frictional forces 0.616 N (ss), 1.228 N (NiTi) for 0.018"×0.025" wires and 0.786 N (ss), 1.174 N (NiTi) for 0.019"×0.025" wires (Tables 2 and 3).

The effect of wire dimensions became clear when comparing the two extremes: passive and active self-ligating systems. For passive self-ligating brackets, there were no significant differences between all

Table 5 Results of dimensional assessment of the bracket slot sizes

Bracket type	Slot width Mean±SD (µm)	Slot height Mean±SD (µm)
Metal Twin Bracket (M-TB)	673.4±23.3	977.1±45.7
Active Ceramic Self Ligating (A-C-SL)	599.1±19.0	566.0±21.0
Active Metal Self Ligating (A-M-SL)	598.2±6.0	484.0±28.5
Passive Metal Self Ligating (P-M-SL)	662.6±17.9	785.0±26.0

dimensions for NiTi wires and only one exception with steel wires: when the highest normal force of 2.5 N was applied. In contrast to this, in tribological systems made of metal active self-ligating brackets and stainless steel wires all dimensions differed significantly from each other (p<0.001). There was a statistically significant order (0.016"×0.022"<0.018"×0.025"<0.019"×0.025") for the wires made of stainless steel. This trend was less clear for the NiTi wire samples, where the dimension 0.016"×0.022" was also lowest and statistically different from the dimensions 0.018"×0.025" and 0.019"×0.025". The larger wire dimensions showed an inconsistent frictional behavior: while metal twin brackets tended towards passive self-ligating brackets, active ceramic self-ligating brackets showed a similar behavior as active metal self-ligating brackets. For stainless steel there was no difference between 0.018"×0.025" and 0.019"×0.025" when active self-ligating brackets made of ceramic are used.

Due to the fact that NiTi and stainless steel wires are used during different clinical phases of the treatment an additional direct statistical comparison between the two materials was considered obsolete. Nevertheless, 0.016"×0.022" NiTi wires combined with active ceramic self-ligating brackets and a normal force F_N=1.0 N produced frictional forces of 0.142 N (Fig. 4), whereas the use of steel wires resulted in a much higher frictional force of 0.358 N. The same was true when larger wire dimensions were used. For 0.019"×0.025" NiTi wires at F_N=1.0 N had a frictional force of 0.356 N, while for stainless steel wires friction was at 0.435 N. Increasing normal force to 2.5 N the difference between NiTi (0.845 N) and stainless steel (1.084 N) was relatively lower for 0.019"×0.025" wires, but still evident.

The results of the dimensional measurements on the archwires are summarized in Table 4, which also gives information about the conformity with the tolerances given in ISO 13996 specification[34]. It is obvious that all measured dimensions are above nominal value with about half of them being within the ISO 13996 tolerances while the other half is above the specified range. It can also be noted that the deviations for NiTi wires are bigger as compared to the stainless steel wires. The dimensional measurements were completed with the data shown in Table 5, which summarizes the data from the microscope measurements of the bracket slots.

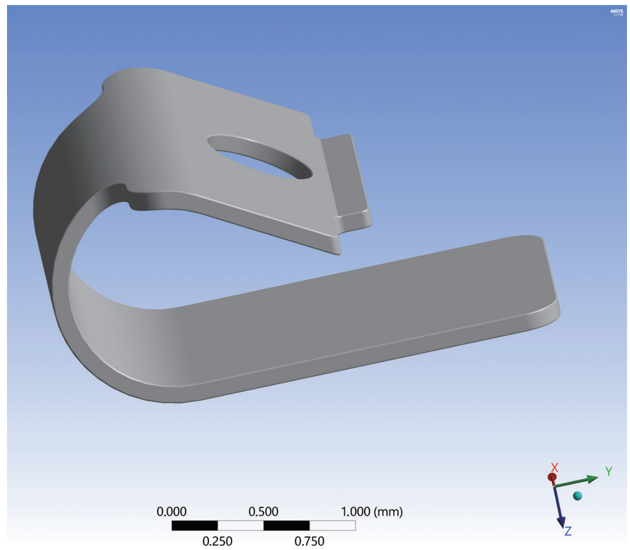

Fig. 5 Exported visualisation of the generated latch model from the investigated active self-ligating bracket, which was used in the frictional tests in the present study.
This model is the basis for the conducted FEM modelling in order to assess the latch force as a function of the archwire height.

It should be noted that the dimensional measurement of the slot height shows higher measurement uncertainty due to the fact that the upper slot limit is rather difficult to identify.

In order to calculate the force generated by the latch mechanism as a function of the applied archwire height, a CAD model has been generated based on the dimensions collected under the stereo microscope in the previous section. This CAD model is shown in Fig. 5.

The SEM surface analysis results are summarized in Fig. 6. The 50× magnification used reveals that all brackets had rounded slot edges so that binding and notching effects should be minimized. Burrs were not present. The surface quality inside the slot was best for the ceramic bracket slot (In-Ovation® C) and least well-polished for the metal twin bracket (Mini Sprint®) while the overall appearance of the surface finish was in all cases acceptable. The slot walls of the Mini Sprint®

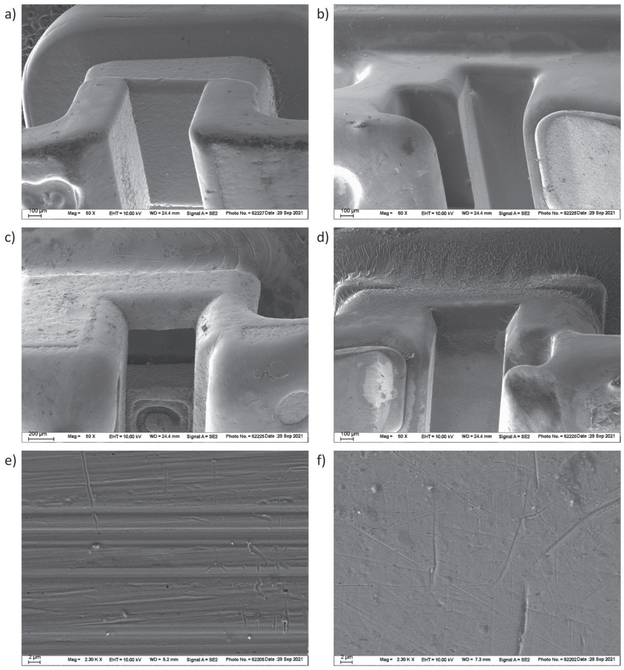

Fig. 6 Overview of the investigated surface qualities on both, brackets as well as arch wires.
a) Conventional stainless steel bracket (Mini Sprint®); b) Ceramic self-ligating bracket (In-Ovation® C); c) Metal passive self-ligating bracket (Damon™ 3MX); d) Metal active self-ligating bracket (BioQuick®); e) Stainless steel arch wire; f) NiTi arch wire

steel bracket offered a slightly less good surface finish as compared to the slot bottom of the same sample as well as compared to the other steel samples (BioQuick® and Damon™ 3MX).

The archwire surfaces of the NiTi and the stainless steel wire showed some significant differences: while the stainless steel wire had significant processing marks on the surface this was not the case for the NiTi wire. In contrast to this, the NiTi wire had higher amount of non-metallic inclusions in its surface. The surface morphology of the NiTi archwire indicated a thorough polishing process leaving minor processing marks with random orientation.

DISCUSSION

Per definition, forces perpendicular to a surface are normal forces and need to be considered in frictional measurements as conducted during the present study. In orthodontics, the force of the ligation has a component acting perpendicularly to the slot floor. As mentioned above[14-16], one can equate the ligation's force as the main contribution to the normal force. Thus, the different types of ligation can result in varying normal forces and their corresponding frictional forces[35-37]. Several studies concluded that an increase in normal force due to tight ligation causes an increase in frictional forces measured[10,13-16,27]. This is in line with the results of the present study. Therefore, only moderately tied stainless steel ligatures or passive self-ligating brackets generate relatively low friction forces favorable in sliding mechanics. Besides other findings, this study showed that active self-ligating brackets performed differently when normal forces were increased or larger wire cross sections were applied. This is most likely due to the active design of the clip used for self-ligating brackets, which pushes the arch into the slot and therefore adds to the total amount of normal force in friction measurement[1]. The combination of variable normal forces and ligation related forces in frictional measurements is unique to the present study.

Data obtained from *in-vitro* studies help to understand frictional force, influencing factors and the relationship to friction and also isolate other variables, such as the effect from ligation mechanism. However, such experiments can simulate clinical situations in orthodontic treatment only to a certain limited extent[25,38]. *In vivo*, a variety of factors during clinical tooth movement such as the angulation and torque between the bracket's slot and the orthodontic wire[2,36] or components of human saliva affect friction[39-43].

Comparing the different kinds of ligation mechanism realized in brackets, passive self-ligating brackets showed the lowest kinetic frictional force, followed by conventional twin brackets and active self-ligating brackets made of ceramics. Active self-ligating brackets produced the highest kinetic friction, especially in combination with large cross-sections of archwires. Similar levels of frictional force in both, passive self-ligating brackets as well as conventional twin brackets, could be due to the play between wire and slot of these two kinds of brackets. It has to be noted that, when conventional twin brackets are ligated with stainless steel wires, the number of twists affects frictional force. Therefore, the force produced by a stainless steel ligature is subjective and the reproducibility of such a ligature is mainly dependent on the orthodontist[44].

Higher friction in active self-ligating brackets may be caused by the spring clip which applies an additional normal force to the archwire[25]. As such, the metal active self-ligating bracket is likely to exert additional pressure onto the wire upon closure. Based on the FEM calculations carried out in the frame of the present study, the amount of clip force generated by the active metal clip is a function of the dimensional height of the applied archwire. While the nominal value of the archwire height can already be considered a criticality for the friction between bracket and wire, the effective value including the dimensional tolerances of the two components in frictional contact may even unintentionally increase the total normal force and thus also the frictional force. Figure 7 shows the effect of the latch clip deflection on the normal force. While it is close to zero for the smallest measured wire size (0.016"×0.022"), the normal force contribution from the clip may increase to about 1.2 N in case of the largest archwire height of 0.019"×0.025" and even up to 1.3 N if the maximum measured wire height

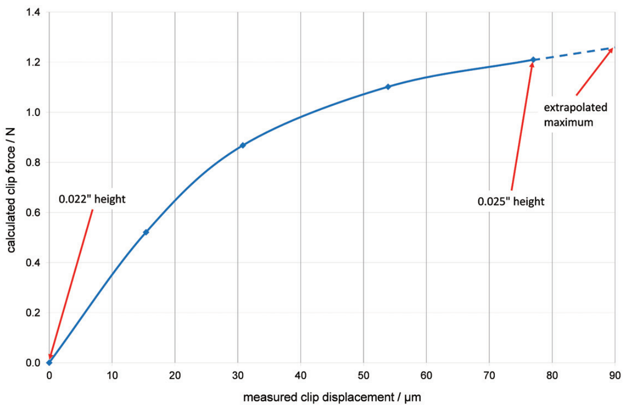

Fig. 7 FEM calculated force-displacement curve for the latch from the active self-ligating metal bracket (BioQuick®, Forestadent).
While a 0.022" wire profile is not actively loaded by the latch mechanism, the calculated maximum of clip force is as much as 1.3 N (profile height of 0.025" plus height maximum tolerance).

is used for the calculations, which raises the total clip deflection to as much as 0.1 mm. This in return means, that for the smaller normal force values applied during this study, the contribution of the latch mechanism is the dominant component.

In contrast, the clip of the metal passive self-ligating bracket creates a closed rectangular tube in the bracket slot and it can be concluded, that this kind of passive slot closure has the least effect on frictional performance of all investigated tribological systems. This is in line with a previous study where passive self-ligating brackets were found to have detrimental effects on torque control due to larger play and the inability to press the archwire against the floor of the slot[45)]. In addition, the apparent play between wire and slot can also influence the frictional force level[8,46)]. If the archwire fills the slot of a bracket and contacts the spring clip or sides without any play, active self-ligating brackets can exert an extra perpendicular force on the wire solely generated by the bracket latch. In contrast, the cover of passive self-ligating brackets has no active force applied to the slot. Thus, the frictional force of passive self-ligating brackets is per se lower than that of active self-ligating brackets[12,15,38,47-51)]. Yet, some authors found no significant difference or observed greater friction in the self-ligating brackets[15,16,52-55)]. Differences between experimental models, including shapes, sizes and prescriptions of self-ligating brackets, as well as clips or caps made of different materials and different companies may explain these inconsistencies. Further studies should test whether these variables have a correlation between each other. In the present study, the archwire was pulled through the bracket while being perfectly aligned and thus under full absence of any tilting as found in clinical tooth movement. This way the effect of the variable "ligature type" was isolated from other effects. The results from the measurement on the slot dimensions show that the slot width is above specification for all samples. However, this finding does not affect the frictional results in our experimental setup because of the alignment procedure between the frictional bodies. Therefore, the play between wire and bracket slot can be neglected.

Wire material is an important factor affecting frictional force. According to the results of earlier investigations, wires made of nickel-titanium alloys produced higher frictional forces than stainless steel wires[8,13,23-27)]. Others observed that there was no significant difference between wires made of stainless steel and nickel titanium[28,29)]. These contradictory findings regarding friction may be attributed to the different forces of ligation employed in various studies and the materials from different suppliers[29)] as well as the surface processing of the archwires which in most above mentioned studies remained unconsidered or was at least unspecified. However, the results of the present investigation showed, that the significantly lower friction provided by wires made of NiTi is most likely either a result of their lower stiffness compared to stainless steel wires or of differences in surface quality. One can conclude from the SEM pictures shown in Fig. 6 that the surface morphology of the stainless steel arch appears to be rougher as compared to the NiTi wire surface. This becomes particularly evident when using wires with large dimensions and is in line with previous studies[10,30,56)], which compared the frictional values produced by these wires under certain contact angulations. Generally, the reason for this is binding angulation. This phenomenon can make the normal force between wire and bracket increase as stiffness increases[10,56)]. Compared to wires made of NiTi, stainless steel wires are less flexible and possess higher stiffness. This may exert greater normal force when the wires are completely ligated into the slot of the bracket. In addition, frictional force produced by NiTi wires was lower than that of stainless steel wires. Therefore, most likely a combination of surface morphology, contact angles and stiffness of orthodontic archwires influenced frictional force values. This is in line with a previous study, where surface treated NiTi wires demonstrated significantly less friction than the non-treated wires[57)]. Still, one has to bear in mind that the materials NiTi and stainless steel are clinically not in competition with each other: NiTi wires are used during the leveling phase, while stainless steel is used during space closure and in the contraction phase.

Finally, there is an obvious difference in dimensions between stainless steel wires and NiTi wires, the latter being larger in dimensions as one can see from Table 4. This finding can be attributed to the unique manufacturing process of NiTi wires in general, which consists of a cold rolling process (rectangular wires are manufactured using a Turk's head machine) followed by a thermal treatment which is carried out in a shape setting tool in order to relief stresses and to "program" the wire to the final arch geometry. However, due to this final heat treatment the material tends to remember its previous shape due to the occurrence of the martensitic transformation in the furnace and thus contracts by a

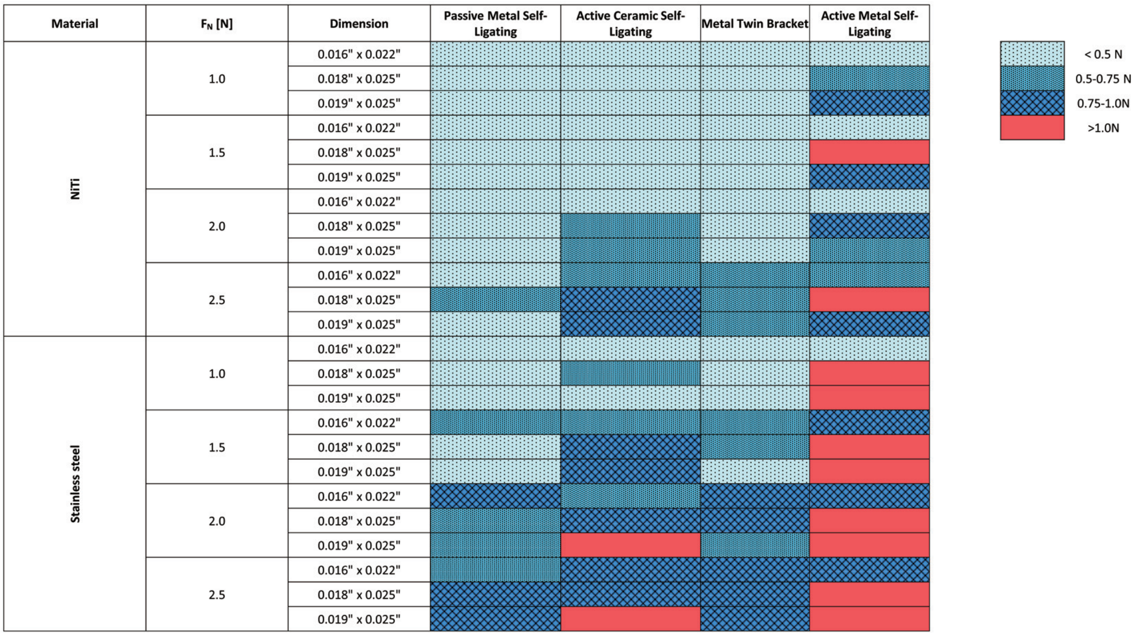

Fig. 8 Visualization of the different levels of friction for the tribological systems tested.
The different levels of friction are color-coded: light blue (<0.5 N), medium blue (0.5–0.75 N), dark blue (0.76–1.0 N), red (>1.0 N).

few percent. As the volume remains constant, the cross section of the wire shows a small increase in dimensions which means that meeting tight tolerances in NiTi wire is more challenging as compared to stainless steel wires which do not undergo a similar shape setting procedure.

In summary, one should carefully choose bracket type, wire material and wire dimension, and consider additionally applied normal forces. This decision has to be made with regard to type of treatment or treatment step, respectively. In Fig. 8 the levels of friction are visualized in four different categories. These categories give the clinician a quick impression, which combinations of bracket type, wire dimensions, wire material and normal force can be used to achieve the desired frictional forces (Fig. 8). If very low frictional forces are needed, combinations with low overall friction are advisable. For example, wires made of NiTi with a low cross section combined with twin brackets with loose wire ligatures or passive self-ligating brackets. Still, to loose a ligature may compromise the three-dimensional control and torque control of the tooth. In contrast, in steps requiring friction, stainless steel wires with large dimensions alongside with active self-ligating brackets are recommendable with the clip exerting extra additional pressure onto the wire. Further studies should address the effect of varying ligatures, such as manual steel ligatures, alastics or differently designed active bracket clips on normal forces.

CONCLUSIONS

From the present *in-vitro* study one can conclude:

1. Frictional forces increase, as normal forces are increased. Thus, only moderately tied ligatures or passive self-ligating brackets generate low friction forces needed during sliding mechanism.
2. Passive self-ligating brackets showed significantly lower frictional forces compared to active self-ligating brackets for almost all wire dimensions tested.
3. Active self-ligating brackets increase the frictional forces due to the contact force between the active latch of the bracket and the archwire surface. FEM simulations indicate forces up to 1.3 N, which has to be added to the applied normal force of 1.0 up to 2.5 N.
4. Stiffer wire materials like stainless steel will contribute to higher friction and therefore to better three-dimensional tooth-movement control.
5. The less well-polished surface of the stainless steel archwire appears to be a factor for increasing frictional forces of stainless steel if compared to NiTi wires.
6. Ligation type —wire combinations should be chosen carefully and appropriately for the intended treatment step.

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