Experimental Investigation of the Biomechanical Properties of a Newly Introduced Self-ligating Bracket

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1. Gutachter: Prof. Dr. rer. nat. Christoph Bourauel

2. Gutachter: PD Dr. Pia-Merete Jervøe-Storm

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Aus der Poliklinik für Zahnärztliche Prothetik, Propädeutik
und Werkstoffwissenschaften der Universität Bonn
Direktor: Prof. Dr. med. dent. H. Stark

- Stiftungsprofessur für Oralmedizinische Technologie -

Prof. Dr. rer. nat. C. Bourauel

To my beloved parents, Niko & Dimitra

and my extraordinary sister, Foteini

Inhaltsverzeichnis

1. Deutsche Zusammenfassung der Dissertation

1.1 Einleitung und Literaturübersicht

Im Rahmen kieferorthopädischer Behandlungsmaßnahmen unterscheidet man im Wesentlichen dentoalveoläre, skelettale und funktionskieferorthopädische Korrekturen. Zur Behandlung dieser orofazialen Abweichungen stehen uns sowohl herausnehmbare als auch festsitzende kieferorthopädische Apparaturen zur Verfügung, welche in aktiver oder passiver Form zur Anwendung kommen können (Kahl-Nieke, 2001). In vielen Fällen stellt die bogengeführte Zahnbewegung die am einfachsten durchzuführende Methode zur Bewegung von Zähnen dar. Als wichtigste technische Elemente der Multiband-Apparaturen sind sicherlich die Brackets und die Drahtbögen anzusehen.

Anfang des vorigen Jahrhunderts wurde von Angle das erste Multibandsystem entwickelt. Seitdem folgten viele Techniken mit dem Ziel, die an den Zähnen angreifenden Kräfte zu reduzieren und so Überlastungsschäden und Wurzelresorptionen zu vermeiden (Kahl-Nieke, 2001). Nach einer Studie von Andrews (1972) wurde die Behandlung mit einer Multiband-Apparatur in verschiedene Phasen eingeteilt. Es wurde zwischen der ersten, der Nivellierungsphase der Behandlung, der Führungs-, Kontraktions-, Justierungs- sowie der Retentionsphase unterschieden, wobei in der letzten Phase das Behandlungsziel bereits erreicht ist und Art sowie Umfang dieser Abschlussbehandlung abhängig von der Ausgangssituation und vom individuellen Entwicklungsstand des Patienten variieren können.

Während der Nivellierungsphase überträgt der in den Bracketslot einligierte Drahtbogen seine Kraft über das Bracket auf den zu bewegenden Zahn und erreicht über einen gewissen Zeitraum einen dreidimensionalen Ausgleich der Slotdifferenzen. Es existieren unterschiedliche Faktoren, die die auftretenden Kraftsysteme während der Nivellierungskorrektur beeinflussen können und sich infolgedessen auf die Nivellierungseffektivität positiv oder negativ auswirken. Insbesondere stellen nach einer Studie von Schumacher et al. (1992) das Bracketdesign, die Ligatur, das Drahtmaterial und die Drahtdimension wichtige potentielle Einflussfaktoren in der Nivellierungsphase dar. Für den optimalen klinischen Einsatz festsitzender Apparaturen ist die Kenntnis, in welcher Art Drahtbogentyp, Bracketbreite, Zahnbogenform, Zahnposition, Kippungsgrad sowie Art der Ligatur, die Korrektur einer Zahnfehlstellung beeinflussen, extrem wichtig (Pandis et al., 2008).

Während der Nivellierungsphase bewegt sich das Bracket entlang des Führungsbogens. Dies verursacht stets eine Reibungskraft zwischen Bogen und Bracket sowie zwischen Bogen und Ligatur. Folglich ist nach der Untersuchung von Schumacher et al. (1992) die während der bogengeführten Zahnbewegung auftretende Reibung ein wesentlicher Faktor, der das Kraftsystem, die Form und auch die Effektivität der Bewegung beeinflussen kann. Jedoch ist die Größe und klinische Auswirkung dieser Reibungskraft weitgehend unbekannt (Andreasen und Quevedo, 1970; Frank et al., 1980; Tidy et al., 1989).

Um bei der Korrektur einer Zahnfehlstellung eine effektive Nivellierung zu erzielen, ist es notwendig, eine entsprechende Kraft auf den Zahn und das umliegende Gewebe auszuüben. Wegen der entstehenden Reibung muss die Größe der applizierten Kraft die Reibungskraft übersteigen, darf jedoch den physiologischen Bereich nicht überschreiten. In der Literatur (Proffit et al., 2007) werden Kräfte zwischen 0,3 und 0,6 N empfohlen, die allerdings nach anderen Studien (Reitan, 1957) bis zu 1,5 N betragen können.

Im Rahmen einer kieferorthopädischen Behandlung ist es auch häufig erforderlich, in den letzten Phasen der Behandlung Winkelfehlstellungen der Oberkieferfrontzähne in orovestibulärer Richtung zu korrigieren. In Abhängigkeit von der Größe der Torsion, der Dimension und der Legierung des Drahtes, dem Spiel des Drahtes im Bracketslot sowie der Verformbarkeit des Brackets bewegt der Vierkantdraht durch die im aktivierten Zustand entstehende Torsionspannung die Zahnwurzel nach vestibulär (Cash et al., 2004; Fischer-Brandis et al., 2000; Harzer et al., 2004). Weiterhin beeinflusst der Interbracketabstand, somit also die freie Drahtlänge zwischen den Brackets in Abhängigkeit von der Bracketbreite, das erzeugte Torquedrehmoment (Creekmoore, 1976; Hemingway et al., 2001; Schudy et al., 1989).

Die Entwicklung sogenannter selbstligierender Brackets geht auf erste Publikationen zur Mitte des letzten Jahrhunderts zurück (Cacciafesta et al., 2003; Keim, 2005). Obwohl diese Brackets schon seit geraumer Zeit auf dem Markt angeboten werden, werden ihre Eigenschaften auch derzeit noch in der Wissenschaft stark kontrovers diskutiert. Erstmals beschrieben wurde diese Bracketart durch Stolzenberg (Stolzenberg, 1935). Selbstligierende Brackets verfügen über spezielle Verschlussmechanismen, die den Draht im Bracketslot verankern können. Verbessertes Torqueverhalten, reduzierte Reibungsverluste, verbesserte Nivellierungseigenschaften, reduzierte Kräfte, geringerer Zeitaufwand, oralhygienische Vorteile, Tragekomfort und auch verbesserte Ästhetik werden häufig als Vorteile selbstligierender Brackets benannt (Harradine, 2008).

1.2 Ziele

Im Rahmen der Multiband/Mulitbracket-Technik waren Neuentwicklungen im Bracketdesign und der Ligierungstechnik stets darauf ausgerichtet, die Übertragung von Kräften und Drehmomenten zwischen Bracket und Bogen zu optimieren sowie dadurch auch eine verbesserte Nivellierungseffektivität und ein ausreichender Torqueübertrag zu erzielen. Heutzutage wird auch für die zahnärztliche Praxis gefordert, dass sie nach evidenzbasierten Prinzipien geführt wird, wobei es besonders wichtig ist, jede kieferorthopädische Apparatur daraufhin zu überprüfen, ob sie in der Lage ist, Zähne effektiv und in vorhersagbarer Art und Weise zu bewegen. (Turpin, 2009).

Anderseits besteht derzeit noch weiterer Klärungsbedarf in Bezug auf die Torquecharakteristik unterschiedlicher Draht/Bracket-Kombinationen. Dies kann insbesondere an der Komplexität der klinischen Situation und an der Vielzahl unterschiedlicher Parameter, die Einfluss auf die Torqueeigenschaften haben, liegen (Harradine, 2003).

Im Rahmen dieser in-vitro-Studie wurde daher untersucht, ob sowohl das Nivellierungsverhalten als auch der Torqueübertrag durch verschiedene Bracketsysteme bzw. durch die Ligierungstechnik beeinflusst werden können. Dabei stellte eine kombinierte Fehlstellung die Ausgangsposition dar, die mittels eines biomechanischen Messgerätes zu korrigieren waren. Dadurch konnten sowohl die Nivellierungseffektivität der Brackets und Bögen, die erzeugten Kräfte, die als Nebenwirkung auftretende Torquebewegung, als auch der Torqueübertrag und das Torquespiel ermittelt werden.

1.3 Material und Methode

Für die Simulation der Nivellierungskorrektur und der Torquebewegung wurden die zu untersuchenden Brackets und Bögen in einem biomechanischen Messsystem (Orthodontisches Meß- und Simulations-System, OMSS) eingebracht (Bourauel et al., 1992;

Drescher et al., 1991). Das OMSS hat sich in einer Reihe vorangegangener vergleichbarer Studien hervorragend für die Untersuchung derartiger Fragestellungen bewährt (Montasser, 2013).

Von Frasaco-Modellen (Frasaco GmbH, Tettnang) eines vollständig bezahnten eugnathen Oberkiefers wurden Kunststoffreplikas (Palavit G 4004, Heraus Kulzer GmbH, Hanau) hergestellt. Der rechte obere Schneidezahn wurde aus dem Modell entfernt, alle übrigen Zähne wurden mit dem zu untersuchenden Bracketsystem nach Herstellervorschrift im ausnivellierten Zustand beklebt. Die so vorbereiteten Modelle dienten der Messung der von verschiedenen Draht/Bracket-Kombinationen erzeugten Kraftsystem in den folgenden unterschiedlichen Situationen: (a) erzeugte Kraftsysteme und Korrekturseffektivität bei einer Fehlstellung des Schneidezahnes von 2 mm nach gingival und 2 mm nach labial; (b) erzeugte Drehmomente und Torquespiels nach einer Rotation des oberen rechten Schneidezahnes im Sinne eines Torque von ± 20 ^{o}.

Hierzu wurden die Modelle so im OMSS integriert, dass der entfernte Schneidezahn durch ein Bracket ersetzt werden konnte, das über einen Hebelarm mit einem Sensor des OMMS verbunden wurde (Montasser, 2013). In dieser Studie wurden drei Bracketsysteme des 0,018"-Slotsystems und ein neuartiges Bracketdesign des 0,016"-Slotsystems untersucht. Die folgenden Brackets wurden untersucht: (1) konventionell ligierendes Kunststoffbracket (Brilliant®, Forestadent, Pforzheim), (2) konventionell ligierendes Metalbracket (Mini Mono®, Forestadent), (3) aktiv selbstligierendes Bracket (Speed™, Strite Industries, Ontario, Canada), (4) passiv selbstligierendes Bracket (Swiss Nonligating Bracket, SNB, Tröster Applications, Magden, Schweiz).

Die Messungen erfolgten in Kombination mit den folgenden Drähten:
(a) für die Nivellierungsmessungen: (1) 0,007" NiTi (Tröster Applications), (2) 0,009" NiTi (Tröster Applications), (3) 0,0135" NiTi (Tröster Applications), (4) 0,016" NiTi (Tröster Applications), (5) 0,016" NiTi-BioStarter® (Forestadent), (6) 0,016" Stahl (Forestadent).
(b) für die Torquemessungen: (1) 0,016" x 0,016" Stahl (Forestadent), (2) 0,016 x 0,016" NiTi-BioTorque® (Forestadent), (3) 0,016" x 0,022" Stahl (Forestadent), (4) 0,016" x 0,022" NiTi-BioTorque® (Forestadent).

Das Einligieren der konventionellen Brackets erfolgte mit Hilfe von Elastikligaturen (Dentaurum). Da NiTi-Drähte ein temperaturabhängiges mechanisches Verhalten aufweisen, war der gesamte Messaufbau in einem Klimaprüfschrank mit einer konstanten Temperatur von 37 °C eingebaut. Vor Beginn jeder Simulationsmessung wurde darauf geachtet, dass die Temperatur 37 (± 1) °C betrug.

Für jede einzelne Draht/Bracket-Kombination wurden fünf Messungen durchgeführt. Die Nivellierungsmessumgen wurden in eine große Zahl von Schritten (bis zu 200 Messzyklen) unterteilt und nach jedem Schritt erfolgte eine erneute Messung der Kraftsysteme mit Neuberechnung der entsprechenden Zahnbewegung. Durch diese zyklische Wiederholung konnte der Veränderung der Kraftsysteme im Verlauf der Zahnbewegung Rechnung getragen werden. Anschließend wurde die Korrektur der Fehlstellung automatisch vorgenommen. Für die Torquemessungen wurden die Rotationen von 20 ° ebenfalls in kleine Schritte, insgesamt jeweils 80, unterteilt. Durch zyklische Wiederholung konnte die Registrierung der erzeugten Drehmomente und des Torquespiels durchgeführt werden. Nach Untersuchung jedes Bracketsystems wurde jeweils ein neuer Drahtbogen eingesetzt.

Für die Auswertung wurden Mittelwerte und Standardabweichungen aller Messwerte ermittelt. Alle Ergebnisse wurden mittels ANOVA auf statistische Signifikanz überprüft.

1.4 Ergebnisse und Diskussion

Ein einheitliches und zufriedenstellendes Bild mit Korrekturwerten bis zu 99 % zeigten die selbstligierenden Brackets. Sowohl die initiale Infraokklusion als auch die vestibuläre Verlagerung wurden durch die selbstligierenden Bracketsysteme mit Werten zwischen 77,9 % und 99,0 % erfolgreich nivelliert. Bei den Nivellierungsergebnissen zeigten sich bei bestimmten Draht/Bracket-Kombinationen scheinbare Überkorrekturen sowohl der vestibulären als auch der vertikalen Verlagerung. Ein Vergleich der Säulendiagramme zur Korrektur der vestibulären Verlagerung und der Torquebewegung zeigte einen deutlichen Zusammenhang der Torquebewegung und der Nivellierungseffektivität in der oro-vestibulären Richtung.

Bezüglich der Kräfte wurden die geringsten Werte bei den selbstligierenden Brackets kombiniert mit dem 0,0135" NiTi-Draht ermittelt. Die hier gemessenen Kräfte waren allerdings allgemein deutlich höher als die entsprechenden in der Literatur empfohlenen Kräfte.

In Bezug auf die Torquemesungen zeigte sich, dass der Einfluss des Drahtes der dominierende Faktor bei der Erzeugung eines Frontzahntorque ist. Insofern ist eine korrekte Auswahl von Drahtquerschnitt und Drahtmaterial die ideale Vorgehensweise, um ein angepasstes und für die erwünschte Bewegung geeignetes Torquedrehmoment zu erzeugen. Der Einfluss der Ligierungsmethode war von geringerer Größe. Zusammengefasst konnte festgestellt werden, dass das Bracketdesign, die Bracketbreite, die freie Drahtlänge, das Draht/Slot-Spiel oder der Grad der Torquefehlstellung einen geringeren Einfluss auf Torquekontrolle und Torquedrehmoment haben.

Insgesamt wurde für alle Ergebnisse ein statistisch signifikanter Unterschied festgestellt, wobei mittels ANOVA ein statistisch signifikanter Einfluss der Variablen Brackettyp ($p<0,05$) und Drahttyp ($p<0,05$) auf die Variablen Kraftniveau während der Nivellierung, Korrektur der Fehlstellung und Torqueübertrag festgestellt werden konnte.

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2. Introduction and literature review

A common characteristic among the scientific society is the constant endeavor to achieve greater conformity between theory and practice. A universally accepted scientific truth, which at any given time best explains a phenomenon is defined as a “paradigm” (Kuhn, 1996). Once a paradigm shift has occurred, new ideas and information are generated, leading to advances in the scientific field (Mohs, 1991). And so goes on the perpetual circle of scientific evolution.

In the rapidly expanding field of dental sciences, in general, and in orthodontics in particular, this progression finds wide application. According to the American Association of Orthodontists, the field of dentistry concerned with the supervision, guidance and correction of the growing and mature dentofacial structures, including those conditions that require tooth movement related by the application of forces and/or the stimulation and redirection of functional forces within the craniofacial complex is formally defined as orthodontics and dentofacial orthopedics (Evans, 1993). The goal of modern orthodontics can be summed up as the creation of a dynamic balance among occlusal relationships, dental and facial esthetics, long-term stability and restoration of the dentition (Graber, 2000). An intellectual challenge for every practitioner displays the use of scientific and artistic foundation for achieving predictable therapeutic outcomes (Proffit and Fields, 2007).

Looking back in history, evidence for existence of any kind of orthodontic appliances can be found at least at 1000 BC in both Greek and Etruscan populations. As dentistry was developing in the 18th and 19th century a number of devices for the regulation of malpositioned teeth were described by various authors and apparently used sporadically by the dentists of that era. Edward H. Angle is considered in the worldwide orthodontic society the father of modern orthodontics by establishing orthodontics as the first dental specialisation, as well as categorizing malocclusion and developing appliances intending to be suitable for treating the types of malocclusion he identified. Among others, he was the first to introduce a standardized fixed appliance (Edgewise Technique). In the upcoming years clinicians have progressively produced modifications and enhancements mainly in order to improve force delivery of the appliances and clinician’s efficiency.

Additionally, recent advances in material science, metallurgy and biomedical engineering have introduced an increasing array of alloys, capable of generating a wide spectrum of mechanical forces. A continuous interaction between orthodontists and engineers has already produced major changes in the design of orthodontic brackets, and the composition of metallic and nonmetallic wires that generate proper orthodontic forces, while controlling factors such as friction and strain (Krishnan and Davidovitch, 2012). This interaction is a fertile ground for the development of new appliances capable of engendering optimal tooth movement, biologically and mechanically, for each patient.

In this direction in the last 15 years self-ligating appliances have captured the interest of clinicians and are increasing in popularity. The concept of the development of bracket systems intends to overcome the limitations of stainless steel and elastomeric ligatures in terms of ergonomics, efficiency, plastic deformation, discoloration, plaque accumulation and friction (Harradine, 2001).

Orthodontic tooth movement results from the application of forces to teeth, through specific appliances that the clinicians select, insert and activate. Teeth and the surrounding supporting tissues respond to these forces with a complex biological reaction that ultimately results in tooth movement through the supporting alveoli. Reducing the unknown factors related to the delivery of treatment can subsequently reduce the variability of its clinical outcome. On the other hand, forces and moments generated during the different stages of an orthodontic treatment display controllable variables. Hence, the understanding of the biomechanical background during every tooth movement plays a prominent role in the accomplishment of the maximal efficiency in the framework of an orthodontic treatment.

2.1 Types of orthodontic tooth movement

Tooth movement can be described and expressed in various ways. According to the different applied moment and forces, in terms of magnitude, direction or point of force application, the essentially infinite array of movements can be categorized into four basic types: tipping, translation, root movement and rotation (Tanne, 1991). A schematic depiction of the main kinds of orthodontic tooth movement is provided in Figure 1.

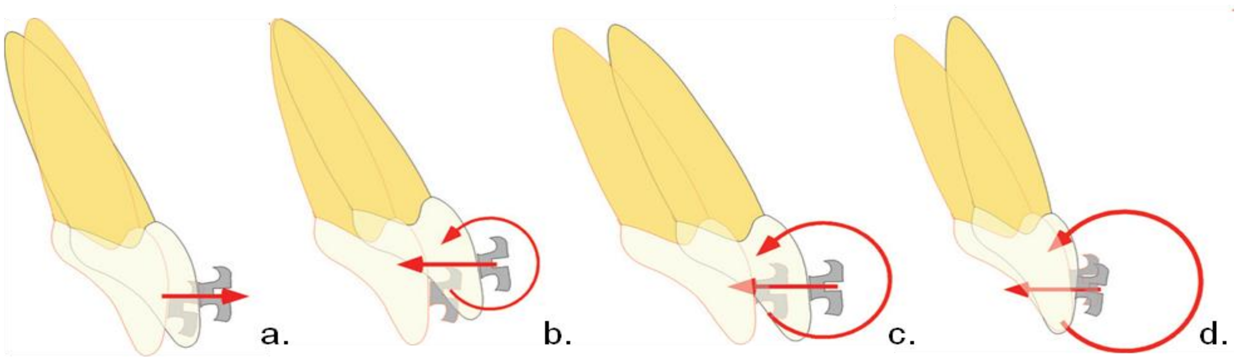

Fig. 1: Main types of tooth movement. (a) Uncontrolled tipping; (b) controlled tipping; (c) translation; (d) torque (Nanda, 2005)

The determination of a tooth movement can be considered by the ratio between the moment created when a force is applied to the crown of a tooth (M_F) and the counterbalancing moment generated by a couple within the bracket (the moment of the couple, M_C). In particular, the correlation of the tooth movement with the moment/force ratio has been presented by Proffit (2000) as follows:

$M_C/M_F=0$	Controlled Tipping
$0<M_C/M_F<1$	Uncontrolled Tipping
$M_C/M_F=1$	Translation
$M_C/M_F>1$	Root movement (Torque)

As far as the rotation is concerned, this type of movement requires a couple of forces. Since no net force acts at the center of resistance, only rotation occurs.

The bodily movement, which in orthodontics is the general definition of a translation, can occur for instance along the tooth’s long axis in an apical direction or in an occlusal direction. Then in the former case we are referring to intrusive movement and in the latter to an extrusive one.

2.1.1 Torque

The term torque presents two different but absolutely interrelated meanings for orthodontists. On the one hand, it can be expressed as buccopalatal root inclination, which can be measured on the lateral cephalometry as the incisor inclination to the maxillary plane. On the other hand, it describes the activation generated by torsion of an archwire in its bracket slot (Rauch, 1959). Depending on the magnitude of the torsion, the size and quality of the wire, the play available for the wire in the bracket slot, the inclination and the deformability of the bracket the archwire moves the root in a palatal direction through the torsional tension induced in the activated state. Incisor torque is often the precondition for a normal interincisal angle, good incisor contact, and sagittal adjustment of the dentition aimed at optimal intercuspation (Harzer et al., 2004).

Numerous authors have attempted to specify the minimum magnitude of torsional moments needed to move the apices of the upper incisor teeth palatally. In particular, Reitan et al. (1964), Jarabak and Fizzel (1972) and Moyers (1973) have estimated the lowest margin of an effective torsional moment to be 0.5 Ncm. Other investigators have evaluated that the value of an adequate torque should not be lower than 1.0-2.0 Ncm (Bantleon and Droschl, 1988; Burstone, 1966; Feldner, 1994; Reitan, 1957).

A very important factor in the expression of the torque capability is regarded to be the interbracket distance, which is determined by tooth and bracket width (McKnight, 1994). In the upper incisor region this distance is evaluated to be approximately 6 mm, as it is stated in the current literature (Feldner, 1994; Holt et al., 1991). Additional it is not only the distance of the bracket placement, but also the vertical positioning of the bracket on the labial surface that plays a prominent role in the torque movement (Dellinger, 1978; Ferguson, 1990; Muchitsch, 1990). Furthermore, the tooth morphology can be also implicated in the adequate torque application (Germane et al., 1989; Miethke 1997). The angle between the long axis of the root and the crown at an upper central incisor may vary widely (Carlsson et al., 1973).

Apart from these factors, the influence of the biomaterials of the fixed appliances should not be underestimated. Only a limited number of studies have dealt with the extent of torque-induced deformation of polycarbonate brackets and with their corresponding torque capacity (Alkire et al., 1997; Dobrin et al., 1975; Feldner, 1994; McKnight et al., 1994). They all suggested, that polycarbonate brackets (reinforced and non-reinforced) expressed a significantly lower torque but more pronounced deformation, when are compared with metal and ceramic attachments. The only exception in the obtained results had been expressed by Dobrin (1975), who claimed that two different non-reinforced polycarbonate brackets displayed pronounced deformation with increasing torque.

2.1.2 Torque loss

In order to achieve useful torque expression, it is necessary to use archwires that engage completely the slot. The size of the slot depends on the technique used. However, the manufacturing process is never absolutely precise, and variations from this nominalvalue do occur. In Germany, the permitted tolerance of bracket slots has been established by the German Institute for Standardization (DIN= Deutsches Institut für Normung e. V.). According to this standardization (DIN 13971-2) a slot tolerance of 0.04 mm can be allowed; i.e the slot width can vary between 0.56 mm and 0.61 mm (equivalent to 0.022 inches and 0.024 inches, respectively). (Sernetz, 2005) Particularly for rectangular archwires, a tolerance of 0.01mm is acceptable. This means that even when acceptable standards are met, there is still a certain amount of play between the archwire and the slot (Ludwig, 2010).

The two basic criteria for assessing the quality of an archwire are its dimensional precision and the execution of edges. It is common sense, that expressing torque becomes increasingly difficult with increased rounding of the archwire edges. Without rounded edges, it would be extremely difficult to insert larger rectangular wires into the bilateral wings. Hence, poor torque expression could be expressed as a result of poor wire engineering and can be associated with the transfer of the torque onto the teeth. Cash et al. (2004) have recently demonstrated a high variability in the shape of the slot which deviated from the rectangular cross-section, as well as in the torque reported and the actual one built into the appliance.

Currently, the evidence available on the relative torque-transmitting efficiency of self-ligating brackets derives from a laboratory study, which suggested that a large torque loss is to be expected in the passive self-ligation comparing it with the active version, where the levels of torque loss appear to be lower (Pandis et al., 2008).

2.2 Alignment and leveling

The idea of dividing orthodontic treatment into stages was emphasized first by Raymond Begg (Begg and Kesling, 1977). The main stages of a comprehensive treatment include:

1. alignment and leveling
2. correction of molar relationship and space closure
3. finishing.

The current study has focused on the biomechanical phenomena, which take place in the first stage of an orthodontic treatment. Thus, this part of the literature review, includes information exclusively for the initial phase of the orthodontic therapy. Initially, the goals of an orthodontist are to bring the teeth into alignment and correct vertical discrepancies by leveling out the arches. Orthodontic movements take effect in various time intervals. Extrusion and elongation of teeth are some of the quickest movements, followed by tipping, which in turn is followed by translation of teeth. On the contrary, torque expression and intrusion of teeth are the slowest observed movements (Ludwig, 2010).

The bracket system, the ligatures, the material and the cross-section of the archwire display factors, which can affect significantly the efficiency of the first phase of the orthodontic treatment. To bring teeth into alignment, a combination of labiolingual and mesiodistal tipping guided by an archwire is needed. Subsequently several important aspects follow this orthodontic mechanotheraphy.

Initial archwires for alignment should provide light, continuous forces to produce the most efficient tipping tooth movement. On the contrary heavy forces are to be avoided. Archwires should be able to move freely within the brackets. For mesiodistal sliding along an archwire, at least 0.02 mm clearance between the archwire and the bracket is needed and 0.04 mm clearance is desirable. For instance, the largest initial archwire that should be used with a 0.018"-slot bracket is 0.016", though 0.014" would be more satisfactory. Rectangular archwires with tight ligatures within the bracket slot should be normally avoided. In these cases, the position of the root apex can be severely affected. The principle is that it is better to tip crowns during initial alignment, rather than displacing the root apices. Superelastic NiTi wires have such low torsional strength that for all practical purposes they cannot torque roots, but the larger wires nevertheless tend to slow the tipping movements needed for alignment. Round wires are preferred for alignment.

2.2.1 Forces generated during alignment and leveling

Optimal forces have been always a controversial topic among the orthodontic scientific society. The initial phase of orthodontic tooth movement involves light and flexible archwires that express relatively low and constant force levels. In 1932, Schwarz postulated that orthodontic forces applied to a tooth should not generate a pressure in the periodontal ligament that is higher than the capillary pressure in it, which is in the range of 0.22 - 0.26 N / cm^2 of root surface, to avoid hyalinization and subsequent root resorption, as well as further complications of the soft tooth supporting tissues. Revising the available literature forces, recommended forces range from 0.3 to 0.6 N (Proffit and Fields, 2007) and can reach up to 1,5 N (Reitan, 1957).

According to von Bohl et al. (2004) high forces in comparison to low ones do not actually result in a faster tooth movement. The difference in the outcome of the movement lies in the response of the supporting soft tissue, as by higher forces more areas of hyalinization are to be observed. In fact, other studies have proved that in the beginning of an orthodontic tooth movement (14 - 28 days from the initial exertion of forces) lighter forces produce a more significant tooth movement (Gonzales et al., 2008). In addition, forces of a higher magnitude produced signs of root resorption.

The amount of force necessary to achieve any particular tooth movement is expressed by the size and the materials of the archwire used, as these directly relate to the clinical force levels exerted on the malaligned teeth. The archwire sizes are increasing subsequently until the phase of the treatment has been completed.

2.3 Orthodontic Wires

Orthodontics distinguishes and differs itself from the other branches of medicine by its widespread use of an array of devices made of almost all the known biomaterials. Orthodontic wires display the active component of the fixed appliances, which generate the biomechanical forces communicated through brackets for tooth movement. Initially, clinicians made attachments using noble metals and their alloys, which were esthetically pleasing and corrosion resistant, but on the other hand, they lacked flexibility and tensile strength. The material that eventually displaced noble metals was stainless steel. Currently, orthodontists principally use wires of four major base metal alloy types: stainless steel, cobalt chromium nickel, nickel titanium, and β-titanium.

2.3.1 NiTi Alloys

The pioneer for the development of nickel-titanium wires for orthodontics was Andreasen (1972) advocating their use in the early 1970s. The first nickel titanium orthodontic wire alloy was marketed as Nitinol (Unitek Corp.) and was developed for the space program (Ni, nickel; Ti, titanium; NOL, Naval Ordnance Laboratory) but has proved very useful in clinical orthodontics due to its exceptional springiness (Graber, 2000).

The nickel titanium wires contain approximately equiatomic proportions of nickel and titanium, and are based upon the intermetallic compound NiTi. Like stainless steel and many other metal alloys, NiTi can exist in more than one form or crystal structure. Austenitic NiTi has a body centered cubic structure that occurs at high temperatures and low stresses. Martensitic NiTi has been reported to have a distorted monoclinic, triclinic, or hexagonal structure and forms at low temperatures and high stresses. For steel and almost all other metals, the phase alteration occurs at a transition temperature of hundreds of degrees (Brantley and Eliades, 2000).

NiTi alloys display two remarkable properties that are unique in dentistry: shape memory and superelasticity. Both these qualifications are related to phase transitions within the NiTi alloy between the martensitic and austenitic forms that take place at a relatively low transition temperature.

Shape memory refers to the ability of the material to remember its original shape after being plastically deformed. The shape memory effect is associated with a reversible martensite-austenite transformation (allotropy), which occurs rapidly by crystallographic twinning at the atomic level. In some cases an intermediate R-phase having a rombohedral structure may form during this transformation process (Proffit, 2000).

There are several important phase transformation temperatures for the NiTi alloys. The transformation range for each of the three structures (austenite, R-phase, martensite) refers to the temperature range for the start and completion of the transformation to the particular structure. For the stress-induced formation of martensite, an additional temperature is defined as the highest temperature at which it is possible to have martensite. Above this temperature stress to form martensite by twinning is greater than the stress for the irreversible movement of dislocations by slip. In a typical application, a certain shape is set while the alloy is maintained at an elevated temperature. When the alloy is cooled below the transition temperature it can be plastically deformed, but when it is heated again the original shape is restored. This property is called thermoelasticity (Bantleon et al. 1989; Brantley and Eliades, 2000; Graber, 2000; Kusy, 1997).

The Nitinol orthodontic wire offered a modulus of elasticity approximately 20 % that of the stainless steel wires, along with a very wide elastic working range. As provided for orthodontic use Nitinol is exceptionally springy and quite strong but has poor formability. In the late 1980s, new nickel titanium wires with an active austenitic grain structure appeared. These wires exhibit the other remarkable property of NiTi alloys – superelasticity - which is manifested by very large reversible strains and a non-elastic stress-strain or force-deflection curve. Part of the unusual nature of superelastic material is that its unloading curve differs from its loading one i.e., the reversibility has an energy loss associated with it, called hysteresis. This means the force that it delivers is not the same as the force applied to activate it. The different loading and unloading curves produce the even more remarkable effect that the force delivered by a NiTi wire can be changed during clinical use merely by releasing and retying it (Bantleon, 1989; Kusy et al., 1997).

The nickel titanium wires have relatively high surface roughness resulting in high archwire-bracket friction. Clinical disadvantages of the nickel titanium orthodontic alloys are that permanent bends cannot readily be placed in the wires and that the wires cannot be soldered. Despite the fact that these kinds of alloys display optimum levels of light force delivery compared to the single-strand wire alloys available, there is concern among some orthodontists about biocompatibility problems associated with release of nickel ions in vivo. Nickel titanium alloys achieve corrosion protection from a passivating surface film of TiO_2. Anodic polarization experiments have shown that the breakdown potentials of these films can vary among alloys suggesting differences in their biocompatibility (Brantley and Eliades, 2000).

2.3.2 Stainless Steel

The family of steels owes its success in the manufacturing of instruments and attachments to combined high mechanical strength, acceptable appearance, and chemical resistance. Stainless steel entered dentistry in 1919, being introduced in Germany by Dr. F. Hauptmeyer. Angle used it in his last year (1930) as ligature wire. By 1937 the value of stainless steel as an orthodontic material had been confirmed (Proffit, 2000). The commonly used stainless steel wire alloys for orthodontic wires are of the “18-8” austenitic type containing approximately 18 % chromium and 8 % nickel. The stainless steels used early in orthodontics were highly resistant to corrosion because of their low carbon and sulfur content and high chromium and nickel content. The chromium in the stainless steel forms a thin, adherent passivating oxide layer that provides corrosion resistance by blocking the diffusion of oxygen to the underlying bulky alloy. Approximately 12-13 wt % chromium is required to impart the necessary corrosion resistance to these alloys (Brantley and Eliades, 2000) Furthermore, chromium, carbon and nickel atoms (and atoms of other metals in the composition) are incorporated into the solid solution formed by the iron atoms. Particularly nickel stabilizes at lower temperatures into a homogeneous and corrosion-resistant austenitic phase. The properties of these steel wires can be controlled over a reasonably wide range by varying the amount of cold working and annealing during manufacture (Kusy et al, 1997).

Research using X-ray diffraction has shown that austenitic stainless steel orthodontic wires may not always possess the single-phase austenitic structure that is based upon a face-centered cubic arrangement of the iron atoms. A two-phase structure was found for some as received stainless steel wires, where the austenitic phase was accompanied by a body-centered cubic martensitic phase. Formation of the martensitic phase results in a substantial reduction in the modulus of elasticity. Extensive cold working can increase the yield strength of these austenitic stainless steels. On the other hand, from a clinical point of view, heat-treating stainless steel aims to minimize breakage rather than to achieve significant increases in resilience. Loss of chromium from the iron solid solution during heating stainless steel depletes the chromium content resulting in a sensitization which leads to a susceptibility to intergranular corrosion (Brantley and Eliades, 2000).

2.3.3 Comparison of the conventional archwires

According to the mechanical properties of the available archwires, it is evident that there is no ideal orthodontic wire alloy. Each of the alloy wires has distinct advantages and disadvantages. Their comparative properties explain why specific wires are preferred for specific clinical applications. Hooke’s law for example, which defines the elastic behavior of materials, applies to all orthodontic wires except superelastic A-NiTi. For everything else, a useful method for comparing two archwires of various materials, size and dimensions is the use of ratios of the major properties (strength, stiffness, range).

Each of the major elastic properties is substantially affected by a change in the geometry of a beam. Both the cross-section and the length of a beam are of great significance in determining its properties. Changes related to size and shape are independent of the material. Generally, as the diameter of a wire decreases, its strength decreases so rapidly that a point is reached at which the strength is no longer adequate for orthodontic purpose. As the diameter increases, its stiffness increases so rapidly that a point is reached at which the wire is simply too stiff to be useful. These upper and lower limits establish the wire sizes useful in orthodontics. By changing the diameter of a beam, no matter how it is supported, its properties are severely affected. When beams of any type made from two sizes of wire are compared, strength changes as a cubic function of the ratio of the two cross-sections; springiness alters as the fourth power of the ratios and range changes as a direct proportion (Graber, 2000; Kusy et al. 2007).

Archwire-bracket friction is high by means of stainless steel because of the relatively rough wire surfaces that arise from the high titanium content. However, clinical studies have not been performed to determine whether the higher sliding friction significantly increases the duration of patient treatment compared to that with stainless steel wires. Andreasen and Morrow (1978) emphasized the benefit of rectangular Nitinol archwires for early treatment stages, where simultaneous rotation, tipping, leveling and torquing could be performed. The superelastic and shape memory nickel titanium wires are particularly useful where large deflections are necessary for extensively malpositioned teeth. Orthodontic appliances fabricated from these wires will generate the most nearly constant forces during the major stages of tooth movement. However, it should be noted that force levels exerted in vivo may not reach the superelastic plateau, and thus the force would not be independent of deflection. A comparative presentation of the mechanical properties of stainless steel and NiTi alloys, as has been summarized by Brantley and Eliades (2000), is provided in the Table 1.

Properties	**Stainless Steel**	**NiTi**
Force delivery	High	Light
Springback	Low	High
Formability	Excellent	Poor
Ease of joining	Can be soldered. Welded joints must be reinforced with solder	Cannot be soldered or welded
Archwire-Bracket Friction	Lower	High
Cost	Low	High
Concern about biocompatibility	Some	Some

Table 1: Comparison of the mechanical properties of stainless steel and Ni-Ti alloys

2.4 Brackets

One of the most important passive components of fixed appliances is the bracket, facilitating merely the attachment of the force producing elements. The evolution of a bondable appliance has demonstrated through the years a remarkable progress curve characterized by extended quiescent periods interspersed with notably active periods.

2.4.1 Self-ligating Brackets

Although the concept of self-ligation was introduced in orthodontics several decades ago, it was only in the last 20 years that these appliances became available in their current form. The first experiments with brackets that fixed the wire into the slot date back to the 1930s. In particular, Stolzenberg (1935) is considered one of the pioneers of self-ligating brackets, by inventing and introducing the Russell attachment (Figure 2). In the upcoming years followed a quiescence period, where many new patents have been presented, but they have been eventually precluded from the clinical practice (Hanson, 1980).

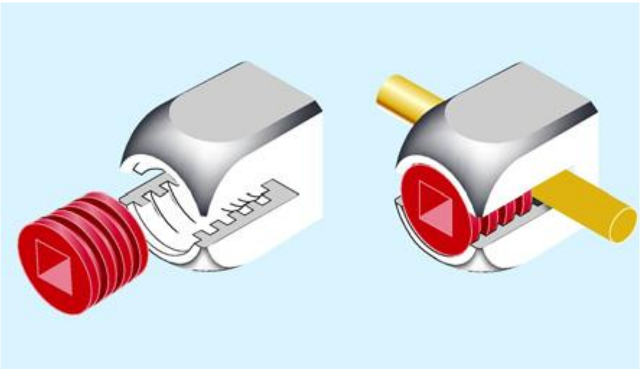

Fig. 2: The Russell attachment. (a) open; (b) closed (Ludwig, 2010)

It was not until the 1970s that interest in the development of self-ligating brackets resurfaced. Since the turn of the century, the pace of development has greatly accelerated with the launch of at least 16 new bracket designs (Eliades and Pandis, 2009).

Like ordinary fixed appliances, a self-ligating bracket consists of a bracket base and a body containing slots and tie wings (Figure 3). The main difference between conventional and self-ligating brackets lies in the way in which the archwire is engaged in the slot.

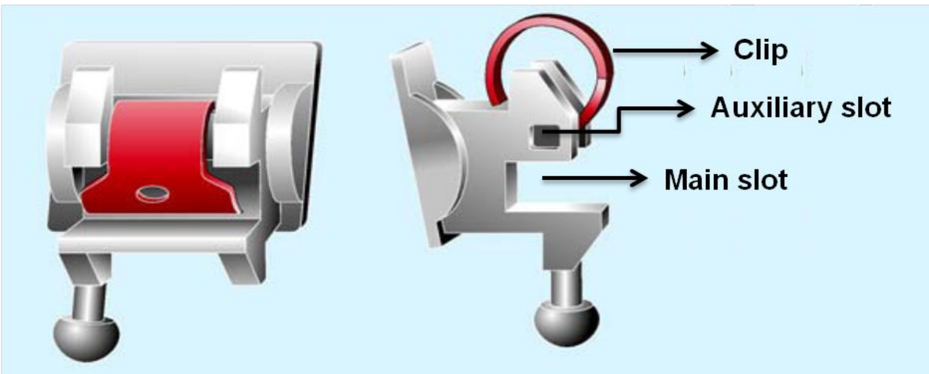

Fig. 3: The general design of a self-ligating bracket. (a) closed; (b) open (modified by Ludwig, 2010)

In self-ligation, the bracket itself contains a clip or other mechanism, which is used instead of either elastic or metallic ligatures. In self-ligation there is a variety of locking mechanisms, which slide in a vertical dimension and can be either rigid (active) or flexible (passive) (Fleming, 2010). The active clip, usually manufactured either from cobalt chromium or nickel titanium, can force the archwire into the bracket slot in a spring-like function. On the other hand, a rigid lid or bolt-like locking mechanism holds the slot shut and subsequently turns the bracket into a tube (passive system) (Ludwig, 2010). In-vitro studies have concluded that the latter self-ligation system displays reduced frictional resistance of the archwire (Fuck et al., 2007). Additionally, deficiencies of the passive clip are associated with inferior rotational and torque control.

Self-ligating brackets can be classified into either tie-wing design, which is accompanied by a locking mechanism, or block design, where any additional modules are not allowed to be attached over the archwire. In the latter design the body is simply used as the retention mechanism for the self-ligating complex (Chen, 2010). Another additional component that is present in the self-ligating bracket design is the auxiliary slot. It enables the use of second force systems, which can be useful in cases of severe derotation, alignment of ectopic teeth, and avoidance of reactive forces during segmental techniques through auxiliary springs (Ludwig, 2010).

2.4.1.1 Proposed core advantages of self-ligating brackets

In the last two decades, a consensus has emerged on the potential core advantages of self-ligation. Considering the properties that are now well documented, as well as those that additional evidence is appearing with increasing frequency, Harradine et al. (2013) have summarized them as follows: faster ligation and archwire removal, secure and full archwire engagement, as well as less chair-side assistance. However, great controversy has been erased amongst researchers as far as the grade of resistance to sliding between bracket and archwire is concerned.

a) Faster ligation

Several consecutive case series studies found out that treatment with self-ligating brackets was quicker, required fewer visits, and resulted in as good or better final alignment and occlusion than treatment with conventional appliances. Nevertheless, other similar studies and all randomized controlled studies to date have presented no difference in these parameters between self-ligating and conventional brackets in various parts of the treatment process (Harradine, 2010). Regarding the conclusions of two recent systematic reviews, from Chen (2010) and Fleming (2010), the evidence supported the view that a treatment with self-ligation results in a less chair time. The reduction in the time needed for ligation holds historically the powerful incentive to develop self-ligating brackets in the era of wire ligation (Shivapuja and Berger, 1994). Several authors have indicated own self-ligating brackets to be better in this respect, with savings of up to 9 minutes per visit compared with wire ligation and approximately 2 minutes compared with elastomeric ligation (Harradine et al., 2010; Maijer et al., 2011; Shivapuja and Berger, 1994; Voudouris, 1997; Turnbull, 2007).

b) Secure full archwire engagement

Absolute security of archwire engagement - as provided by a molar tube - would inherently reduce treatment inefficiencies compared with conventional ligation. Today most self-ligating brackets have mechanisms to deliver this advantage and would be definitely ensure full engagement of all archwires and eliminate the need to regain control of the teeth when full engagement is lost (Harradine, 2001). Furthermore this property according to a recent study from Mezomo et al. (2011) can result in an ameliorating rotational control, as has been observed in cases of canine retraction.

c) Friction

Friction plays a prominent role in the mechanotheraphy of tooth movement. It has been estimated that about half of the force applied to teeth is lost due to frictional resistance (Drescher 1989; 1990). Lower resistance sliding has always gained the scientific interest of many investigators and for this reason in the literature can be revealed a wide range of studies conducted under this perspective. In a comparative assessment of the frictional forces generated between conventionally ligating and self-ligating brackets, the superiority of the latest advancement has been stated by many studies. However many of the obtained results could be easily characterized as overestimated or at least disputable. Binding is regarded to be an important contributor to the expression of friction, which unfortunately cannot be taken into consideration in the framework of an in-vitro investigation. Despite the fact, that has been recently constantly claimed, that self-ligation promotes a decrease in friction, according to the conclusions reached in the study of Thorstenson and Kusy (2001) the contribution to resistance to sliding from the lowered friction of self-ligating brackets is irrelevant when archwire activation causes binding. Moreover, through evaluating the available relevant literature can be stated that round wires contribute in the reduction of frictional forces in sliding mechanics. On the contrary, the biomechanical properties of larger rectangular archwires have not been clearly described and further evidence is still required (Ehsani, 2009).

Summing up, despite the wide range of available studies, no final conclusions in course of the investigation of the influence of self-ligation on the expressed friction can be drawn. The lack of standardization between the tests and the versatile study designs use result in an incapability to transfer the findings of in-vitro studies into clinical scenarios, which are far more complex.

3. Aim of the study

Orthodontic tooth movement is the result of a combination of biological and biomechanical phenomena, which occur under the application of specific forces by means of orthodontic appliances. The complexity and variability associated with biologic systems encourages clinical precision in the application of any stimulus. Reducing the unknown factors related to the delivery of forces can reduce the variability of the treatment's outcome.

Both the behavior of elastic materials and the mechanical factors, that control and influence the effectiveness of the applied forces, must be thoroughly considered in the design of an optimal orthodontic system. In addition, self-ligating brackets have gained lately a broad acceptation in the scientific society and have been extensively investigated. Nevertheless, in these days of multifaceted versatile brackets, the available evidence concerning the potential forces applied to the teeth by the various combinations of self-ligating mechanisms and archwires is limited. In most of the cases the forces derive from simple experimental configurations that do not register force variation as a function of the direction of displacement (Berger, 1990; 1994). As has been previously referred factors including archwire cross-section, bracket slot dimensions, variation in the extend of crowding (Pandis, 2007), biomaterials used, active and passive self-ligation might modulate the forces transmitted to teeth (Iwasaki, 2003).

Under the perspective of the biomechanical principles applied to the orthodontic mechanotheraphy, an in-vitro study was designed and carried out, presenting the following main objectives:

- Comparative investigation of the forces generated during the initial stages of complex orthodontic tooth malalignment correction with various bracket-archwire combinations, by means of an experimental biomechanical set-up.

- Assessment of torque effectiveness in the sagittal plane during later stages of a simulated orthodontic tooth movement by utilizing diverse bracket designs combined with a variety of rectangular archwires.

- Comparative evaluation of potential side-effects manifested by utilizing specific bracket-wire combinations.

Based on these objectives, this experimental investigation intended to provide answers particularly to the following questions:

- Which bracket-archwire combination seems to be the most effective in the phase of alignment?
- Do the self-ligating brackets display better biomechanical properties compared with the conventional ones?
- What are the differences in the behavior of passive and active self-ligating brackets under the same applied mechanical conditions?
- Which bracket system manifests the greatest amount of torque?
- How does the dimension of the archwire's cross section affect the torque effectiveness of brackets?

4. Material and methods

In this chapter, firstly, all the materials that have been used for the purposes of the current in-vitro investigation will be presented successively. Afterwards, the functional principles of the measuring device, orthodontic measurement and simulation system (OMSS), used in order to perform the experimental part of the present study will be displayed.

4.1 Selection of brackets

In order to investigate the alignment capability of fixed appliances, a newly introduced 0.016 inch twin slot bracket (Swiss Nonligating Bracket/ SNB) (Figure 4), as well as 3 different kinds of 0.018 inch slot brackets (Speed, Mini mono, Brilliant) (Figure 5) have been used. Exactly the same bracket systems have been utilized in order to assess their torque efficiency with various archwire combinations. The brackets that have been used, accompanied by their qualities, are presented in the Table 2.

Brackets	**Ligation/ Locking**	**Material**	**Manufacturer**
Brilliant	Conventional	Polyoxymethylene	Forestadent, Pforzheim, Germany
Mini Mono	Conventional	Stainless steel	Forestadent,
SPEED	Self-ligating/ Active System	Stainless steel, NiTi Clip	Strite Industries, Ontario, Canada
Swiss Nonligating Bracket (SNB)	Self-ligating/ Passive System	Polyetheretherketon, Stainless Steel Clip	Tröster Applications, Magden, Switzerland

Table 2: A list of the investigated brackets

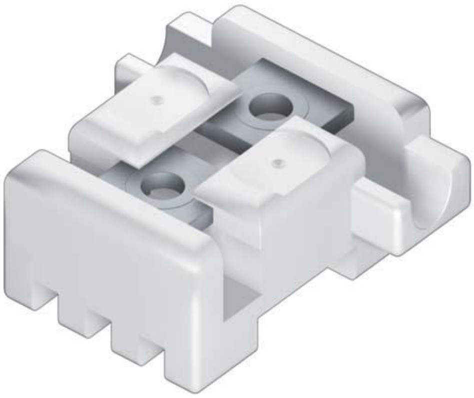

Fig. 4: The Swiss Nonligating Bracket, with its two 0.016 inch slots and passive slides

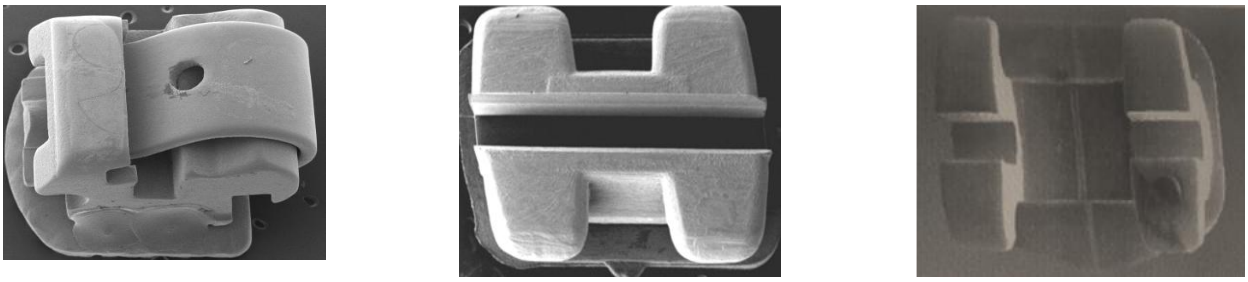

Fig. 5: Scanning electron micrographs of the investigated brackets: (a.) Speed (Morina et al., 2008); (b.) Brilliant (Morina et al., 2008); (c.) Mini Mono (Harzer et al., 2004)

4.2 Selection of archwires

A variety of archwire combinations has been used in order to evaluate the amount of the correction of a complex malalignment and the torque efficiency of the above mentioned bracket systems. These archwires are differentiated basically according to both their material and cross-section dimensions. In the Table 3 all the archwires that have been used during the proceeded experiments in order to evaluate the corresponding alignment and torque capabilities are presented analytically.

Archwire	**Cross section Dimension (mm)**	**Cross section Shape**	**Material**	**Manufacturer**
0.007″	0.18	●	NiTi	Tröster Applications, Magden, Switzerland
0.009″	0.23	●	NiTi	
0.013″	0.33	●	NiTi	
0.016″	0.40	●	NiTi	
0.016″	0.40	●	NiTi/ BioStarter	Forestadent, Pforzheim, Germany
0.016″	0.40	●	Stainless Steel	
0.016″ x 0.016″	0.41 x 0.41	■	NiTi/ BioTorque	
0.016″ x 0.022″	0.41 x 0.56	▬	NiTi BioTorque	
0.016″ x 0.016″	0.41 x 0.41	■	Stainless Steel	
0.016″ x 0.022″	0.41 x 0.56	▬	Stainless Steel	

Table 3: A list of all the investigated archwires together with their cross-section shape and manufacturer

The archwires into the slot of all the conventional brackets have been ligated by means of elastomeric ligatures (Ø 1.3 mm, Dentaurum, Pforzheim, Germany).

4.3 Orthodontic Measurement and Simulation System (OMSS)

The description of a sensor, which enabled the registration of uniplanar forces and moments two-dimensionally, was first presented by Burstone et al. (1976) and later by Solonche et al. (1977). An advanced type of these initially developed experimental devices has been demonstrated by Hershey et al. (1981). Furthermore, similar attempts have been presented by other investigators (Helms, 1977; Fritz and Wurll, 1986), evaluating simultaneously more than one parameter of the biomechanical systems.

Bourauel et al. (1990) and Drescher et al. (1991a; 1991b) based on patents held by Schmieder (P 35 16 234 1; P 35 08 610.6) constructed in 1990 the measuring device of simulation of an orthodontic movement (Orthodontic Measuring and Simulating System, OMSS). It presents a further evolution of all the above described devices, displaying an appropriate size and measuring range, evaluating subsequently the force systems during a simulated orthodontic tooth movement in all spatial directions (Table 4). By revising the relevant literature multiple investigators can be found, who have used this particular biomechanical set-up. For this reason, a brief description of the main functional properties of the OMSS follows below.

Simultaneous registration of three forces and three torques High mechanical stiffness	
Forces: measuring range (resolution)	±15.00 (0.02) N
Torques: measuring range (resolution)	±450.0 (0.5) Nmm
Maximum error in linearity	0.3 %
Maximum error due to cross-talk	1.8 %
Physical dimensions	60 x 60 x 60 mm^3

Table 4: Specifics of force-torque transducers (Bourauel et al., 1992)

4.3.1 Description of OMSS

OMSS comprises of the following main components (Fig. 6):

- a personal computer running the control software
- two three-dimensional positioning tables (Fig. 7)
- two force-torque transducers (Fig. 7)
- two single-board microcomputers
- two stepping motor controller for the movement of the positioning tables in six axes
- temperature controlled chamber

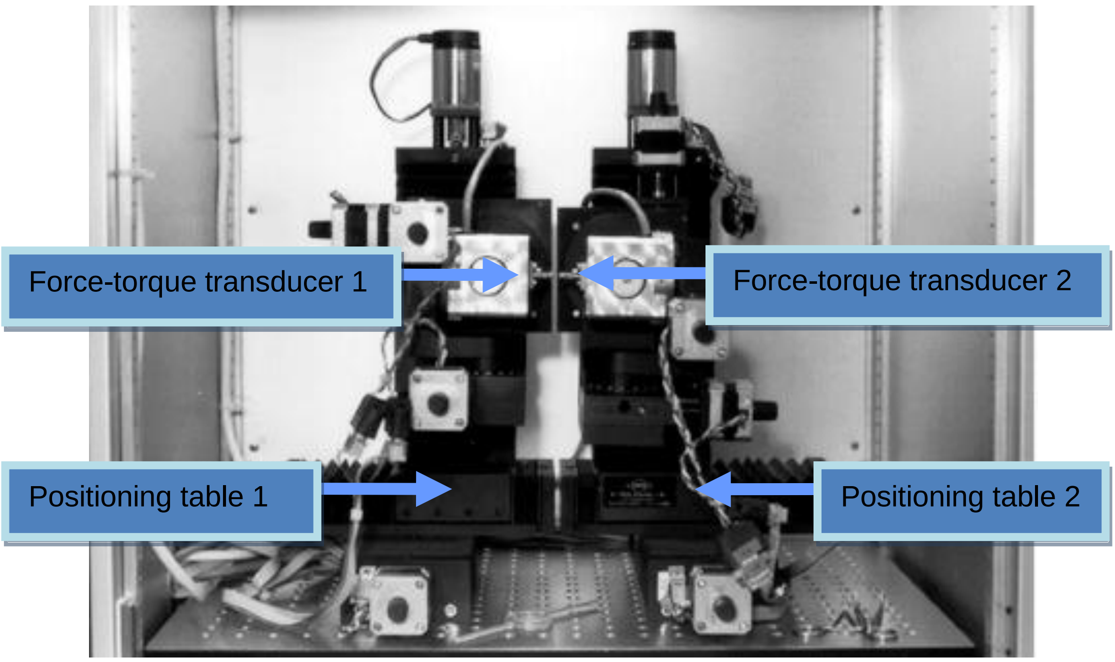

Fig. 6: Construction of OMSS, built up in a temperature controlled chamber

Each positioning table and force-torque transducer is divided into a mechanical and electronic part. The control of each subsystem is performed by its own single-board microcomputer. The whole system of five microcomputers is connected by EPNet, a simple local area network based on the RS232C interface. The major part of OMSS is the force-torque transducer, which measures simultaneously forces and torques in all spatial directions.

Each sensor displays an all-aluminium construction consisting of a central element with four axes. As spatial mobility is required, both sensors are mounted on a set of three linear and rotational stages, which are driven by stepping motors. Commands from the main computer are sent to a single-board microcomputer via serial data network. The micro-computer translates the commands into signals controlling the stepping motor boosters. Due to the sine-cosine amplifiers the motors run extremely smoothly performing small steps (i.e 20 microsteps per motorstep) (Bourauel, 1990). It is worth referring, that in the current study has been used only one force-torque transducer, as for the purpose of the current investigation the malpositioning of only one tooth has been required.

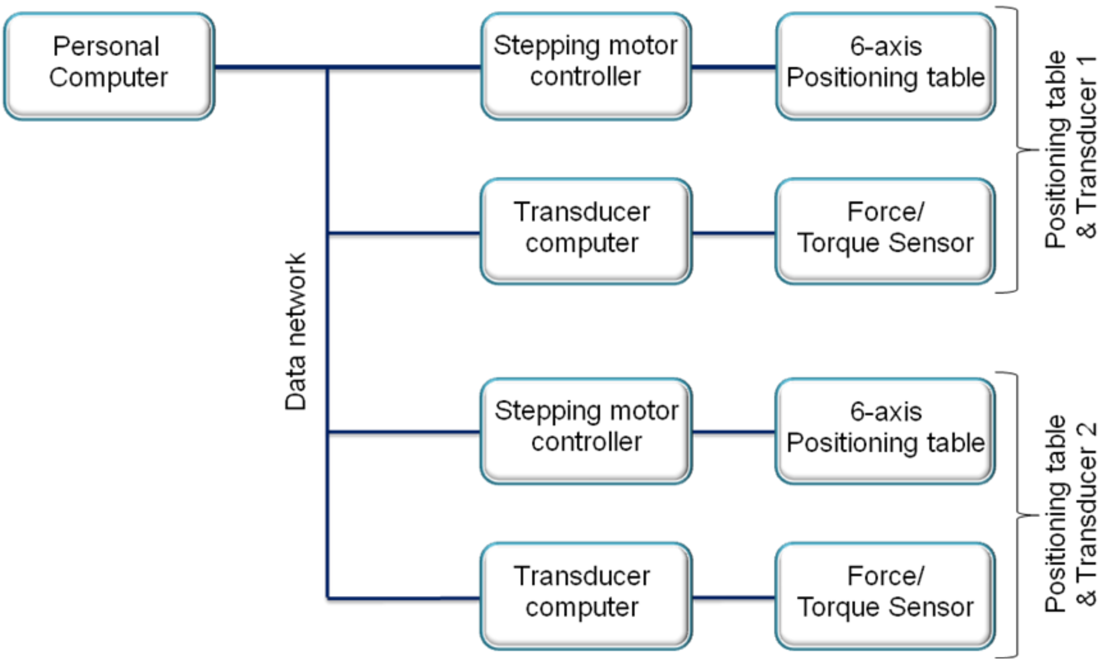

Fig. 7: Schematic presentation of the main components of the OMSS

Additionally, the first part of our experimental investigation includes mainly the application of thermosensitive NiTi alloys in order to control the alignment effectiveness. As has been previously demonstrated the alloy NiTi displays a distinct dependency of elastic properties on temperature. Hence, in order to avoid the extreme alterations in the pseudoelastic behavior of the thermosensitive archwire, the experiments had to be performed in a temperature-controlled chamber at a stable temperature of 37 ^{o}C.

The force-torque vectors generated by the appliance were measured by the sensor and transferred to the computer running the OMSS program, which calculates force-torque vectors acting on the center of resistance of a tooth. A single force applied to the center of resistance causes pure translational movements, whereas force systems in orthodontics always result in additional rotation. Eventually, the resulting vectors for the movements of teeth were calculated through a mathematical model.

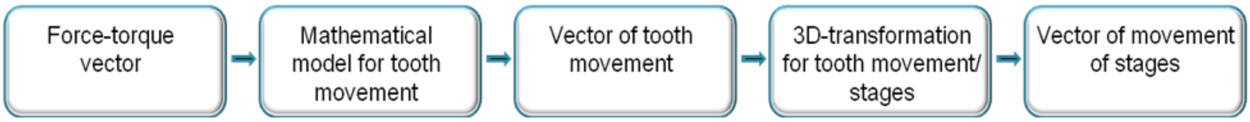

Fig. 8: Symbolic flow-chart of measurement and simulation process

4.3.2 Description of the experiment

This experimental study can be divided into two parts. Initially the measurements concerning the alignment capabilities have been performed and subsequently the torque properties. In order to simulate a human dental arch, four resin replicas (Palavit G 4004, Heraus Kulzer, Hanau) were constructed from a Frasaco model (Frasaco, Tettnang) of a normally aligned maxillary arch. The right central incisor was removed from the resin model to allow placement of a sensor of the experimental setup. Afterwards, brackets were bonded from second premolar to second premolar on the resin models with a cyanoacrylate adhesive. Another bracket appropriate for the right upper incisor was attached to a force sensor of the OMSS. A jig was used in order to standardize the bonding process of this specific bracket (Fig. 9).

Then, the constructed model was mounted on the OMSS table, and the corresponding archwire was applied and engaged in the bracket slot. Due to the fact that the free movement of the archwire in the slot throughout the test is essential, all the ligated wires were not cinched back. At last, the bracket holder with the bonded bracket of the right central incisor was fixed to the left force sensor. The sensor was then adjusted so that the bracket was in a force equilibrium position. The self-ligating brackets in this study were used in the close position. On the other hand, the conventional brackets were tied by means of a needle holder with elastomeric ligatures. Before each measurement, in order to allow a reproducible amount of stress relaxation to occur, the elastomeric ligatures used had a 3-minute waiting period, as recommended by Henao (2004). The whole assembly now simulated the originally aligned arch.

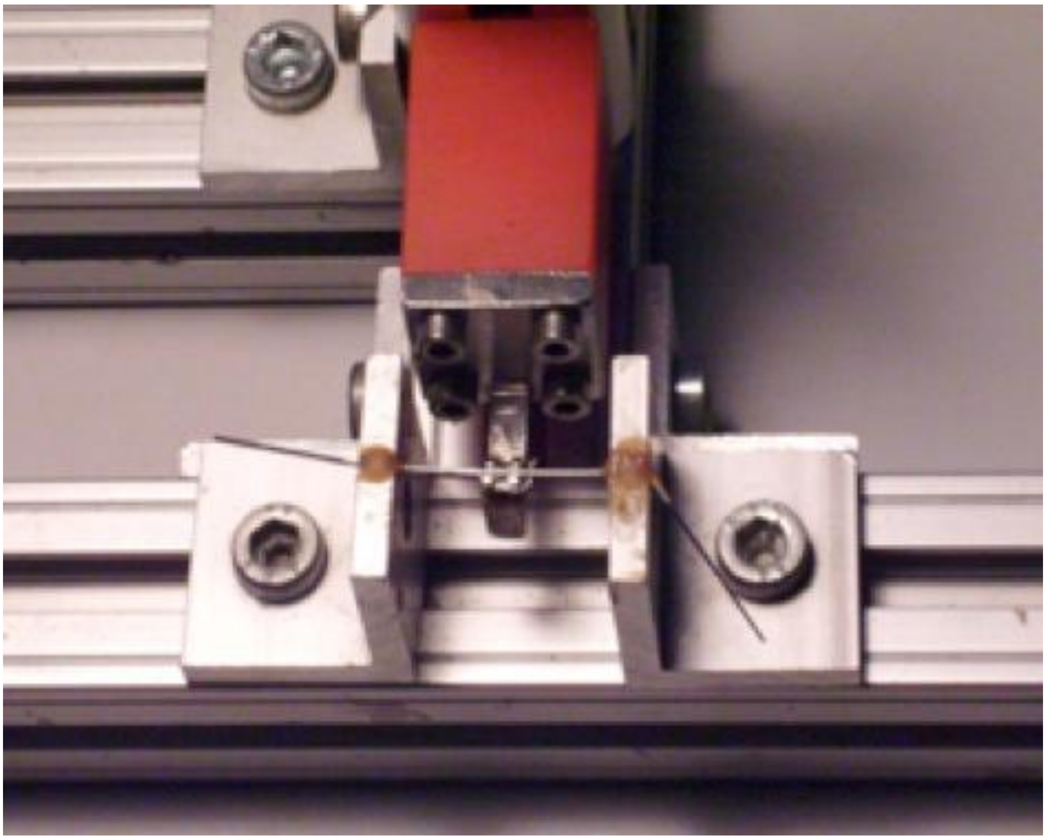

Fig. 9: A right upper incisor bracket bonded on a jig in order to standardize the placement of this bracket for each separate measurement (Fansa, 2009)

At this point the OMSS was set to move the central incisor in order to simulate a specific malocclusion. In particular for the investigation of the alignment the sensor was moved from its initial position 2 mm gingivally and 2 mm labially (Fig. 10) and then should move back to its initial position in small increments. After each step a registration of the generated forces and the proceeded translation was performed. This cyclic repetition of the measurement has taken place 200 times, before one whole experimental investigation was finished. Nevertheless, the process was interrupted in occasions that the predetermined final position was reached, the forces were below a certain threshold or a maximum number of simulation cycles was exceeded. Additionally, this coupled movement coincided with the global x-axis and z-axis, respectively. The initial bracket position was reproduced before its measurement.

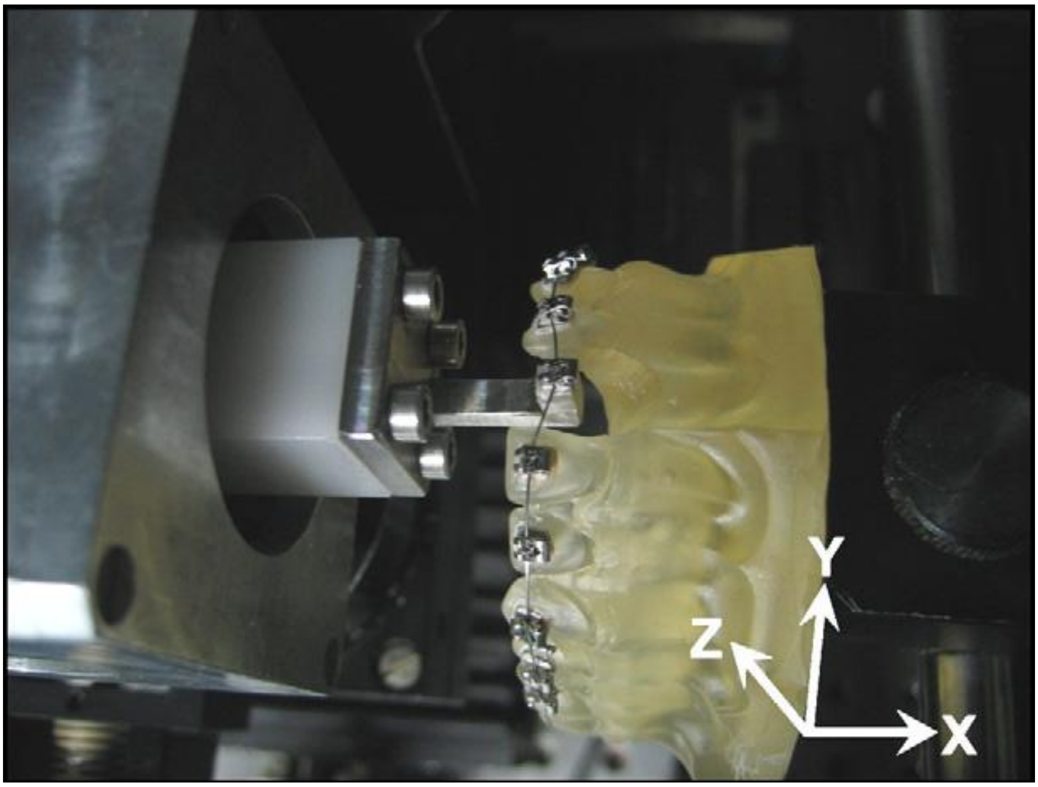

Fig. 10: The resin replica of a Frasaco model mounted in the setup. The upper maxillary right incisor was moved 2 mm gingivally and 2 mm labially. The global coordinate system of the setup is indicated on the lower right.

Each measurement of a bracket-archwire combination was repeated five times. During the measurements with the NiTi wires the temperature was kept constant at 37 ^{o}C (± 1 ^{o}C). At the end of each measurement the maximum absolute values of the forces generated by the bracket displacement in the 2 directions was recorded directly with the OMSS software.

Concerning the evaluation of torque the teeth were moved in a different way. A simulated rotational movement around the long axis of the tooth was applied gradually, ranging from -20 ^{o} (palatinal root torque) to +20 ^{o} (labial root torque) in steps of 0.25 ^{o} increments. Each of the bracket-archwire simulations was performed five times at the entire range of movement, with a newly inserted wire. During the measurements with the NiTi archwires the experiments were performed under a stable temperature of 37 ^{o}C.

During the whole experimental process there had been used the same brackets, which were bonded only once on the resin replicas.

4.4 Data evaluation

After each completed alignment measurement, results concerning the maximum absolute values of the generated forces were registered by means of the experimental setup. Additionally, the alignment effectiveness was calculated in relation to the initial and the final position of the tooth, and could be expressed through the following equation:

Alignment effectiveness (%) = (FP-IP)/FP*100 (%)

FP = Final position

IP = Initial position

In the current investigation the behavior of bracket – archwire combinations in a complex malposition was evaluated. Therefore, in the next chapter the results will be presented separately for the x-axis and the z-axis. The additional rotation around the y-axis that has been observed displays a side-effect of the corresponding biomechanical orthodontic system and will be also described.

Likewise by completion of the simulated rotational movement the values of moments generated were measured, recorded and saved in the computer using the OMSS software. Moreover, through the obtained data the torque play of each archwire in the corresponding slot was evaluated.

4.5 Statistical analysis

Descriptive statistics including means, maximum and minimum values, as well as standard deviation were calculated for each bracket/archwire combination. These data were statistically analyzed with two way analysis of variance (ANOVA) followed by one way analysis of variance. In particular in the observing data of the alignment measurements the forces are the dependent variables, whereas the brackets are the independent ones. As far as the torque measurements are concerned, the absolute values of the moments generated from the rotational simulation were analyzed with a two-way analysis of variance procedure. In this occasion, the moments are serving as the dependent variable, while the bracket and the direction of rotation were set as the predictor variable. Group differences were further analyzed with the Tukey post-hoc comparisons test with the family error rate set at .05. All statistical analyses were performed with the software SPSS Version 20.

5. Results

This section presents analytically all the results of the main issues of this study gathered by the experiments performed with the above mentioned measurement device. In particular, in the first part, all the obtained data concerning the alignment effectiveness will be presented. Curves and descriptive statistics illustrating the forces generated, the rotation and the grade of the correction of the given complex malalignment will be displayed. In the second part, the results representing the torsional behavior of the brackets by inserting rectangular archwires in their slot will be described. To demonstrate the obtained results, pool-diagrams were constructed for a comparative assessment of the investigated properties.

5.1 Alignment and leveling

Each experiment has been fulfilled after 200 measurement cycles. The force, translation and rotation of the tooth have been recorded in each one of these measurement cycles in the three 3 spatial directions by means of OMSS. In the following diagrams, the behaviour of the biomechanical properties expressed indicatively by the SNB combined with a specific archwire will be presented.

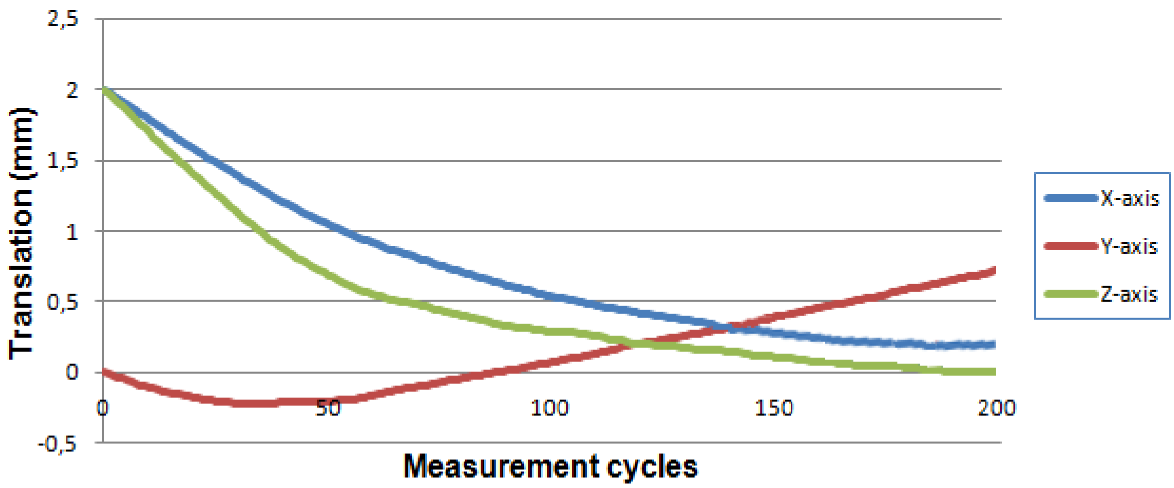

Fig. 11: Translation curve of a tooth under the application of a 0.016" NiTi / Tröster Appl. in the slot of a SNB in terms of a complete measurement with the OMSS

The correction of the alignment can be evaluated by observing the progress of the translation in the diverse axes during one measurement. According to the above provided diagram (Fig. 11), the tooth has reached its optimal position in the z-axis (labio-oral movement) after 200 measurement cycles. Nevertheless, the tooth did not respond exactly in the same way in the x-axis, as the correction appears to be incomplete (92.5 %) at the end of the measurement. The tooth has started its movement from an intruded position of 2 mm and ended up still in an intruded position of 0.2 mm.

This deflection by the end of the measurement can be attributed to the generated rotation around the y-axis, which is considered to be a side effect of the stimulated orthodontic movement. The developed rotation around the y-axis can be defined as a torsional movement. In the following diagram the progress of the corresponding rotation during a whole alignment measurement is manifested.

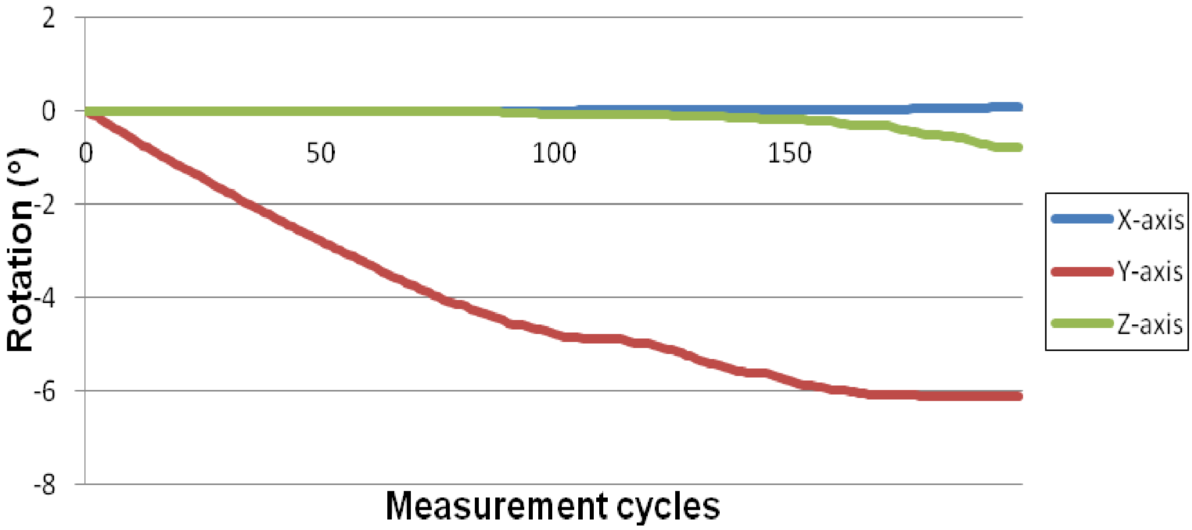

Fig. 12: Rotation curve of a tooth under the application of a 0.016" NiTi / Tröster Appl. in the slot of a SNB in terms of a complete measurement with the OMSS

In this specific occasion, according to the diagram (Fig. 12), torque has reached the value of 6.1 °.

The third diagram (Fig. 13) provides us information concerning the three-dimensionally evolution of the generated forces during the proceeded movement.

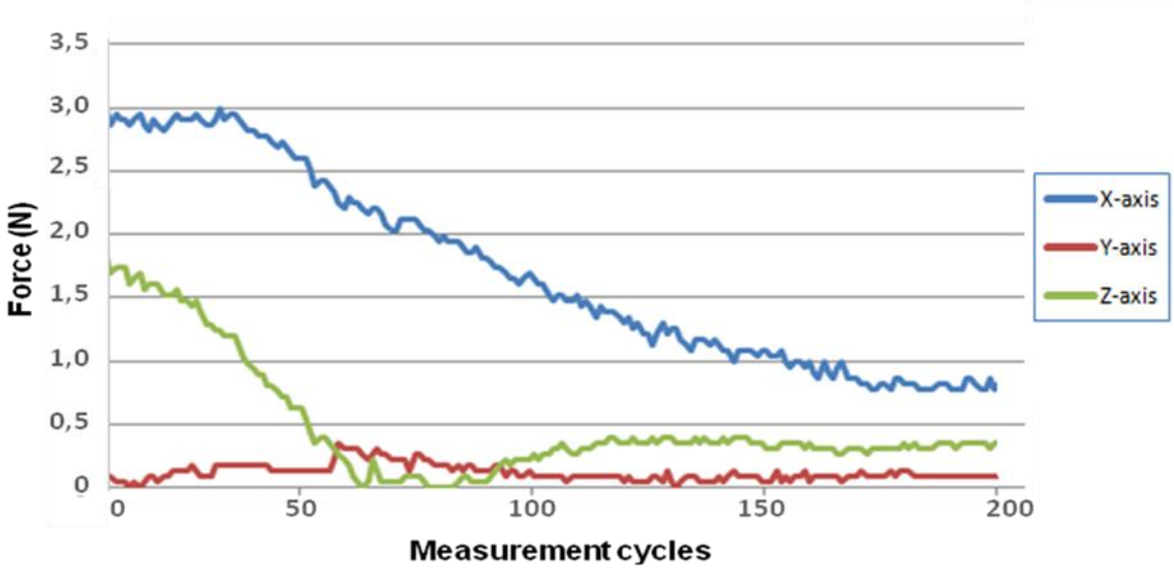

Fig. 13: Force curve of a tooth under the application of a 0.016" NiTi / Tröster Appl. in the slot of a SNB in terms of a complete measurement with the OMSS

By means of inserting a 0.016" NiTi archwire the recorded maximal forces had approached values of 3.0 and 1.7 N for the x- and the z-axis, respectively. In the z-axis the values are flattened after the first 50 measurement cycles, whereas in the x-axis a smooth gradual decrease of the forces' magnitude is observed.

Moreover, the maximal forces applied during the alignment process have been registered through the corresponding force curves. Furthermore, the degree of torque movement has been estimated through the rotation curves. Subsequently, descriptive statistics of the evaluated parameter related to the alignment capabilities will be provided.

5.1.1 Correction (%) of the malalignment

The leveling effectiveness was evaluated on the basis of the achieved correction of the deflection at the end of the simulated movement and expressed in percentages. To demonstrate a detailed presentation of the obtained results bar graphs referring separately to each measurement have been constructed and will be presented below (Fig. 14-19).

Subsequently, to facilitate a comparative evaluation of the obtained results pool graphs referring to the achieved correction (%) at the end of the 200nd measurement cycle in x-axis (Fig. 20) and z-axis (Fig. 21) will be displayed.

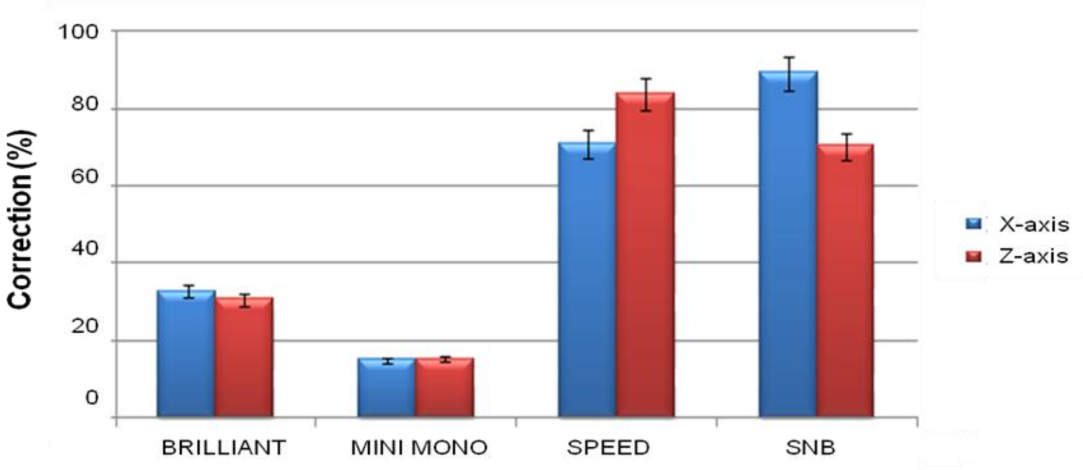

Fig. 14: Achieved correction (%) at the end of the 200^{nd} measurement cycle by combining the various bracket designs with a 0.007" NiTi archwire. The standard deviations are also integrated in the graph.

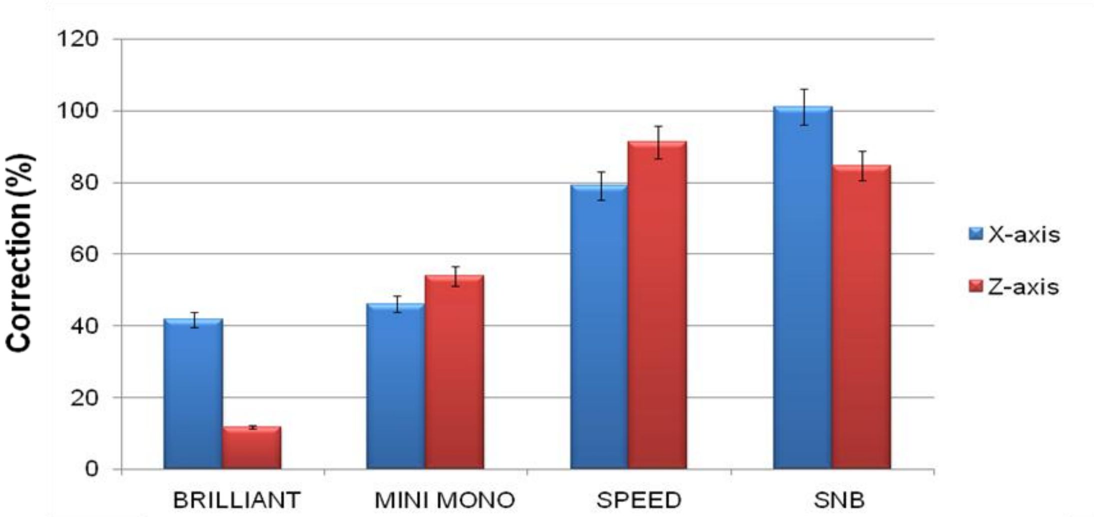

Fig. 15: Achieved correction (%) at the end of the 200^{nd} measurement cycle by combining the various bracket designs with a 0.009" NiTi archwire. The standard deviations are also integrated in the graph.

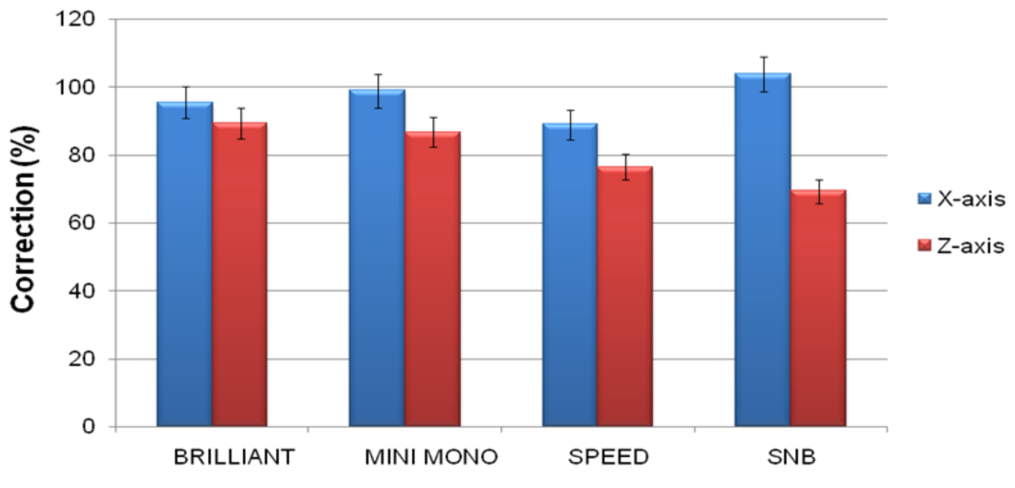

Fig. 16: Achieved correction (%) at the end of the 200nd measurement cycle by combining the various bracket designs with a 0.0135" NiTi archwire. The standard deviations are also integrated in the graph.

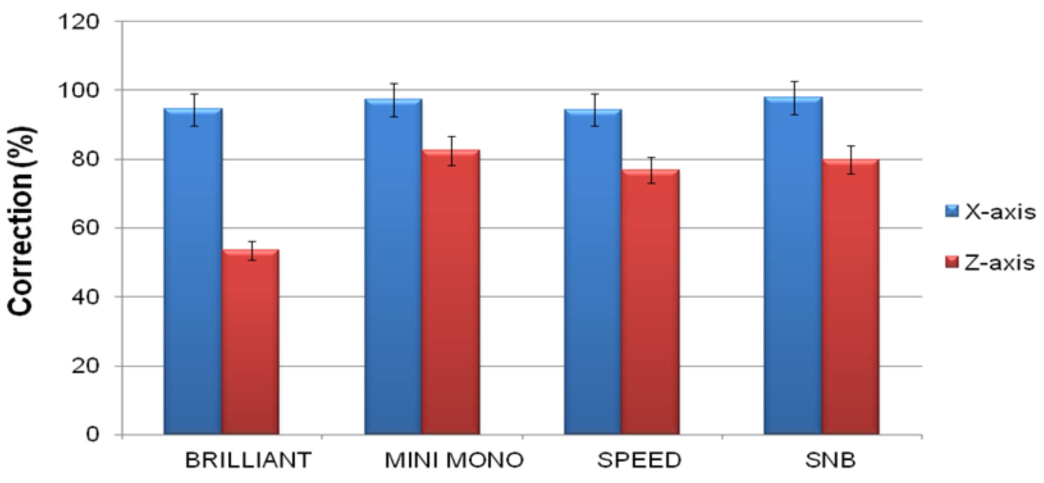

Fig. 17: Achieved correction (%) at the end of the 200nd measurement cycle by combining the various bracket designs with a 0.016" NiTi / Tröster Applications archwire. The standard deviations are also integrated in the graph.

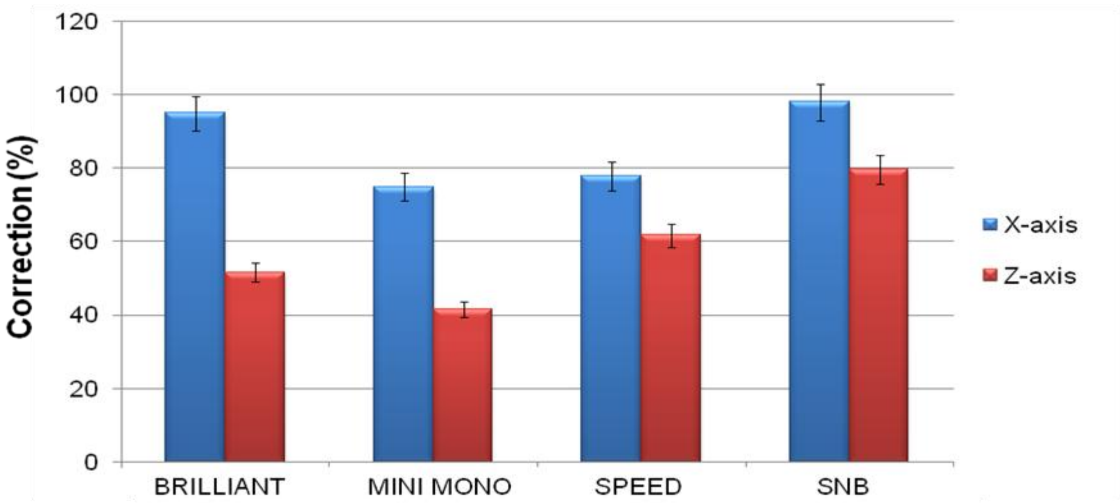

Fig. 18: Achieved correction (%) at the end of the 200nd measurement cycle by combining the various bracket designs with a 0.016" NiTi / BioStarter archwire. The standard deviations are also integrated in the graph.

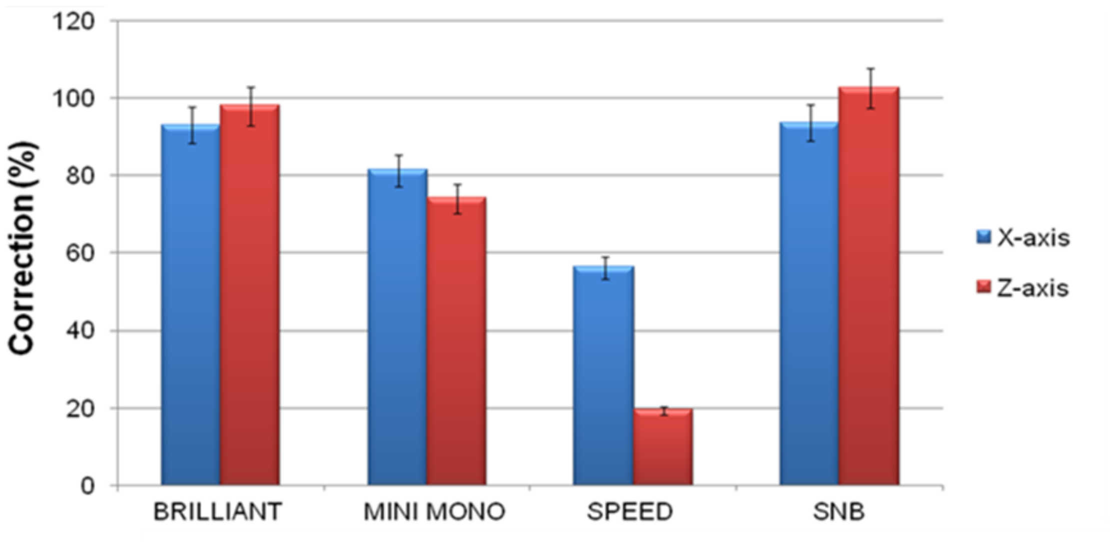

Fig. 19: Achieved correction (%) at the end of the 200nd measurement cycle by combining the various bracket designs with a 0.016" stainless steel archwire. The standard deviations are also integrated in the graph.

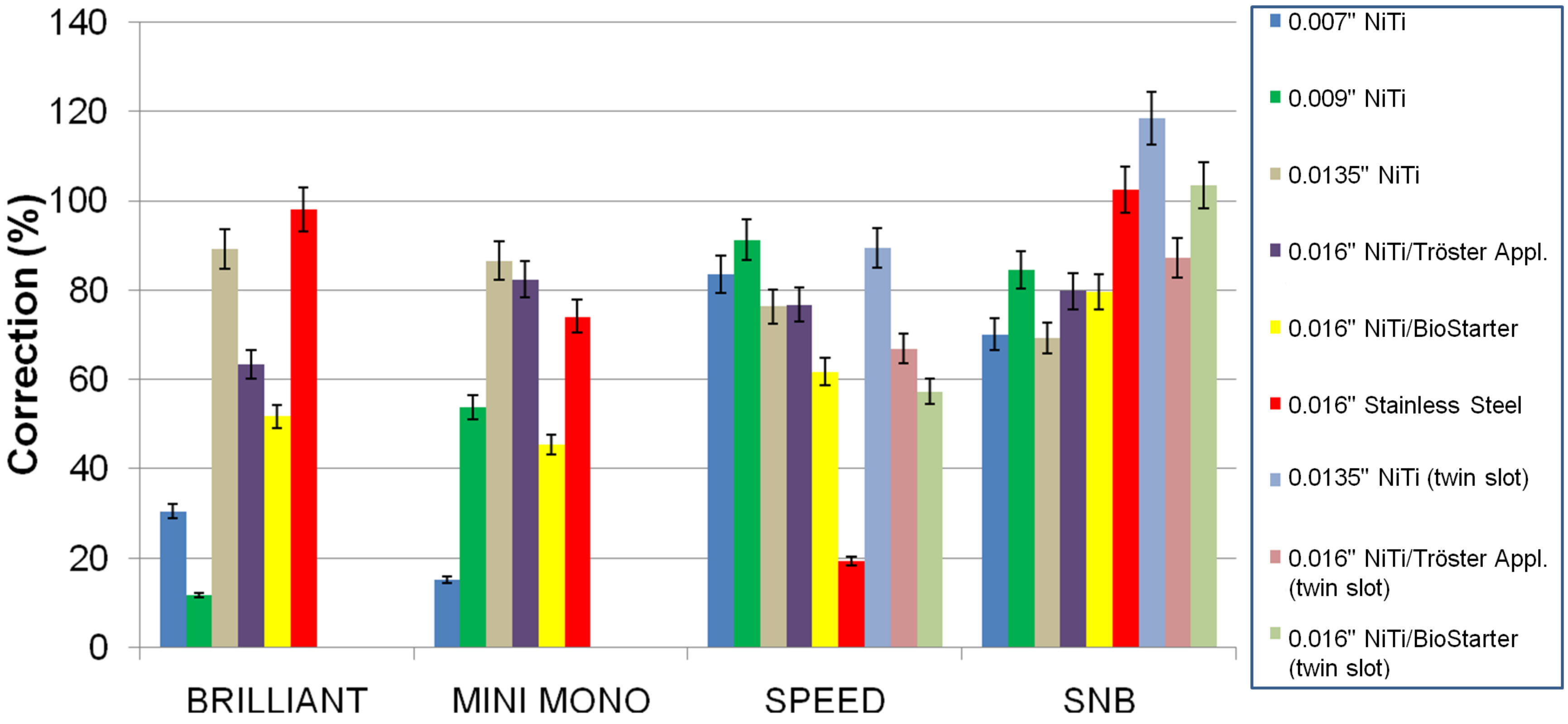

Fig. 20: Pool graph of the achieved correction (%) at the end of the 200nd measurement cycle in the x-axis (infraocclusion) by combining all various bracket designs with all investigated archwires. The last three bars of the self-ligation brackets indicate results by a simultaneous engagement of both slots with the mentioned wires. The standard deviations are also integrated in the graph.

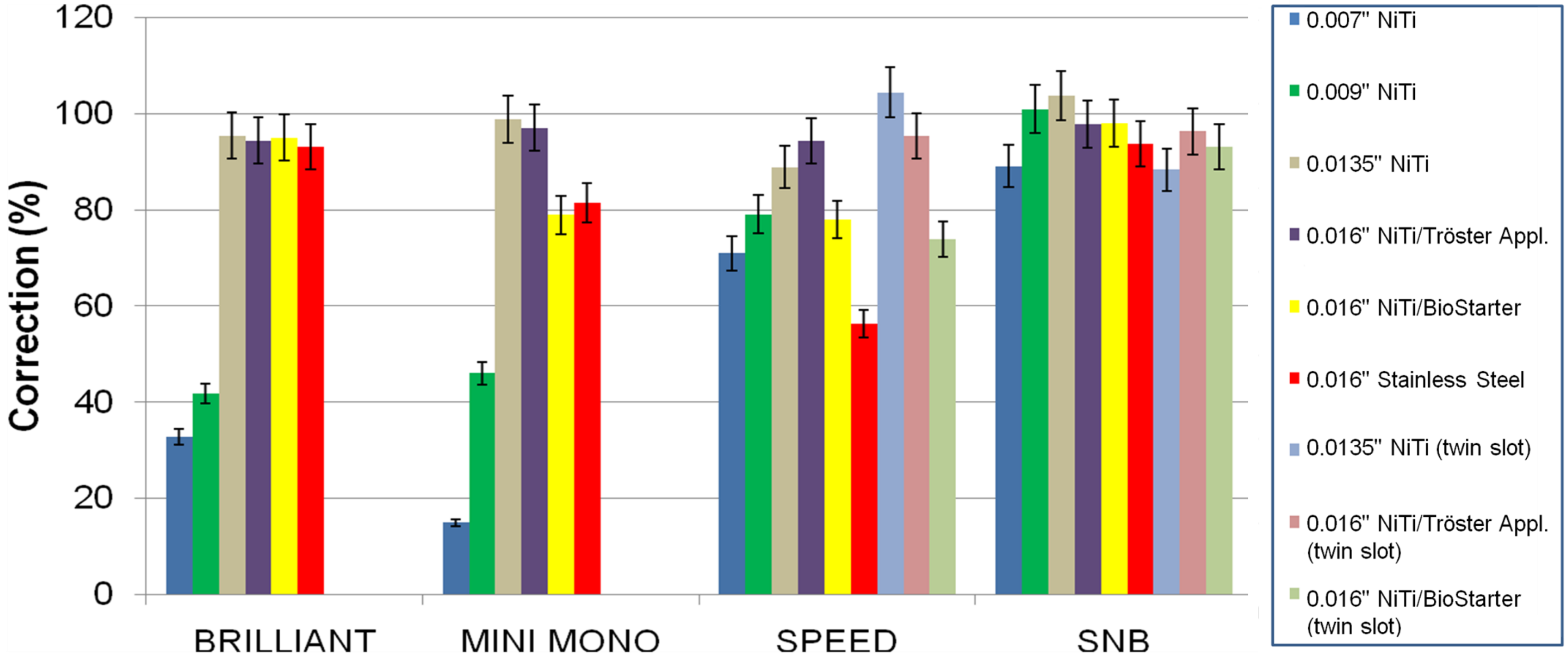

Fig. 21: Pool graph of the achieved correction (%) at the end of the 200nd measurement cycle in the z-axis (labio-lingual movement) by combining all various bracket designs with all investigated archwires. The last three bars each of the self-ligation brackets indicate results by a simultaneous engagement of both slots with the mentioned wires. The standard deviations are also integrated in the graph.

According to the above presented diagrams better alignment capabilities on the whole are exhibited in the z-axis (labioorally), rather than in the x-axis (infraocclusion). The most effective correction on the vertical plane (x-axis) has been achieved with the SNB by insertion in their slot of a 0.016" NiTi / BioStarter archwire (98.44 %). On the contrary, the lowest values concerning this parameter are recorded by combining Mini Mono brackets with a 0.007" NiTi wire (15.0 %). Likewise, as far as the best alignment effectiveness in the labiolingual direction (z-axis) is concerned, SNB combined with a 0.016" NiTi / Tröster Applications archwire display the best response to the orthodontic movement. However, the lowest values of correction in the z-axis were registered by insertion of a 0.009" NiTi archwire in the slot of Brilliant brackets.

An interesting finding of the obtained results concerns the overcorrection achieved in both directions. This phenomenon is particularly related with the self-ligating brackets. SNB demonstrate an overcorrection in the x-axis when they are combined with 0.0135" NiTi and a 0.016" NiTi / BioStarter in both slots, as well as a 0.016" stainless steel. However, this phenomenon in the z-axis is not only manifested by the SNB but by the Speed brackets, too. By engagement of a double 0.0135" NiTi archwire in the corresponding slots the levels of correction that are produced are higher than 100 %.

5.1.2 Generated rotation during the phase of alignment

As has been previously mentioned, the grade of correction of a malocclusion is in interdependence with the generated rotation around the y-axis. According to the aggregated results, which are presented in the graphic below (Fig. 22), self-ligating brackets display on the whole a more prevalent rotational movement greater than 5 ^{o}. This phenomenon is more consistent when the wire is engaged simultaneously in both slots of self-ligating brackets. In particular, the values of the torsional movement during the alignment and leveling phase regarding the 0.007" NiTi and 0.009" NiTi archwires combined with all bracket types display relatively low values, precluding though SNB, where the presented torque is in total higher than 5 ^{o}. The extremely high torque value (18 ^{o}) recorded during the engagement of 0.016" stainless steel wire in the slot of Brilliant brackets presents an exceptional finding of this parameter.

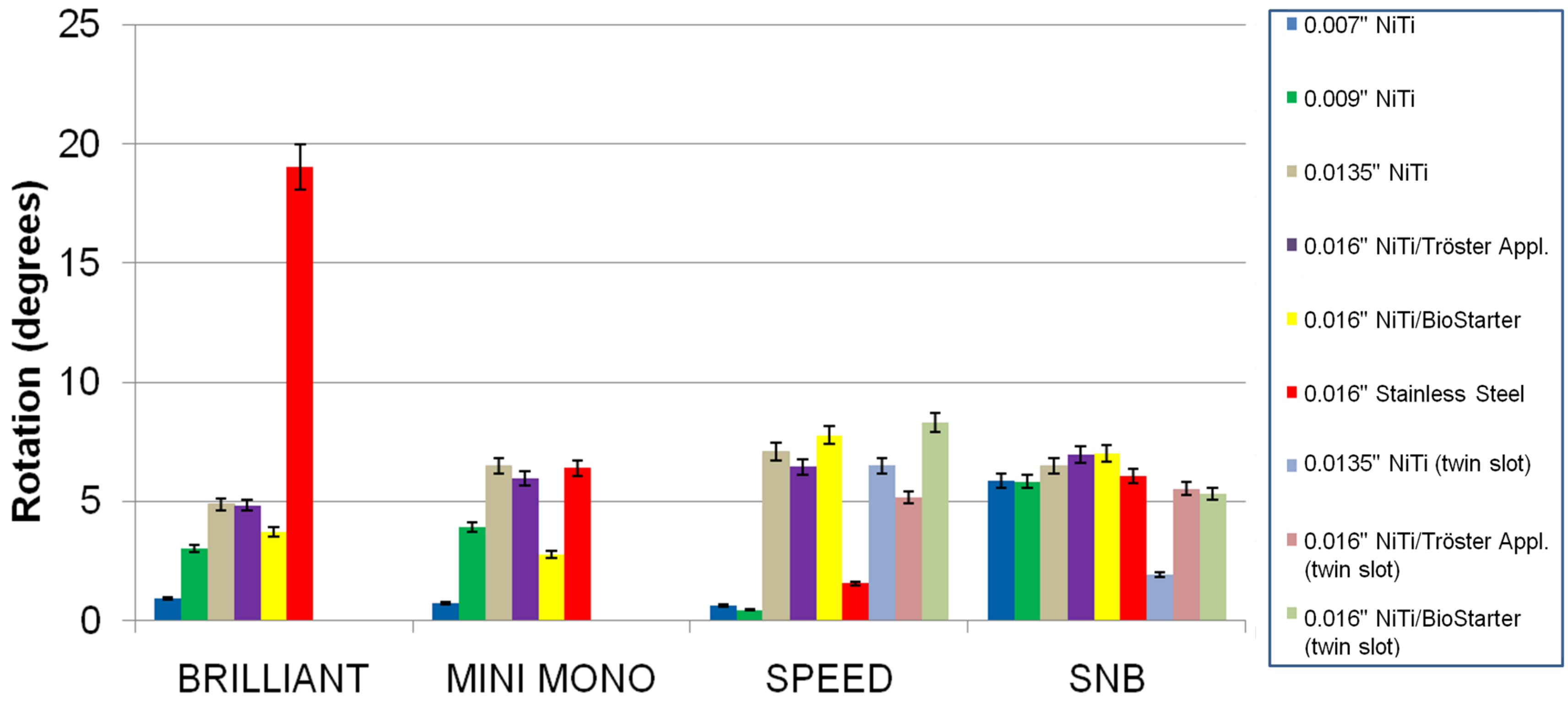

Fig. 22: Pool graph of the generated rotation at the end of the 200nd measurement cycle by combining all various bracket designs with all investigated archwires. The last three bars each of the self-ligation brackets indicate results by an engagement simultaneously of both slots with the mentioned wires. The standard deviations are also integrated in the graphic.

5.1.3 Maximal Forces

In this section, the data concerning the maximal forces generated at the beginning of the alignment and leveling phase both in the x- and the z-axis for each bracket/archwire complex will be presented analytically by means of diagrams (Fig. 23 – 33).

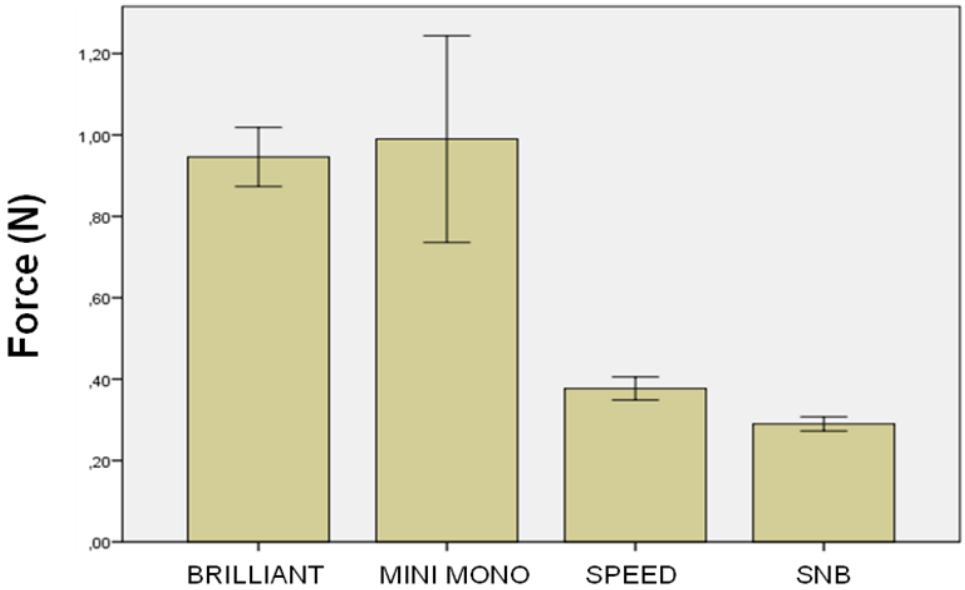

Fig. 23: Maximal forces generated at the beginning of the alignment and leveling phase in the x-axis by application of a 0.007" NiTi wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

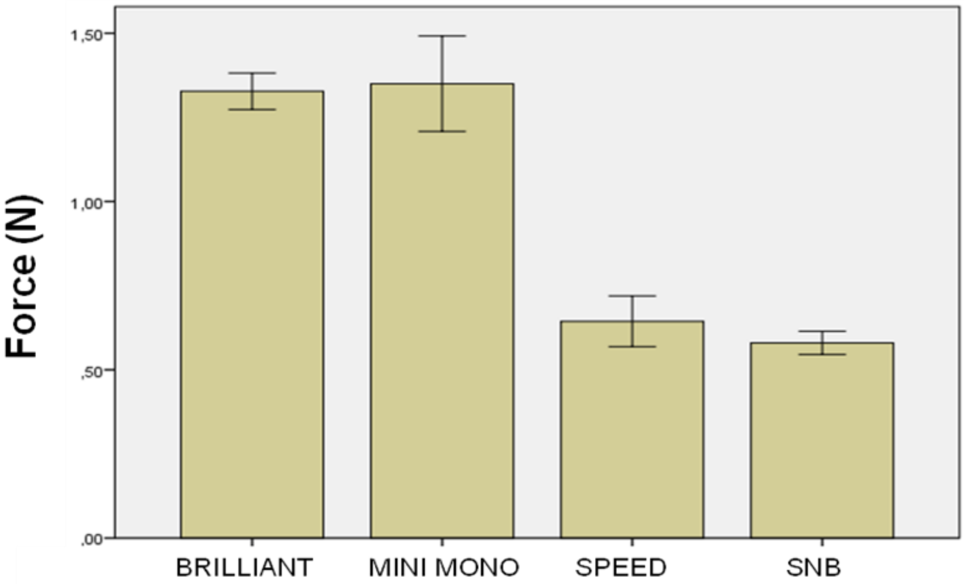

Fig. 24: Maximal forces generated at the beginning of the alignment and leveling phase in the x-axis by application of a 0.009" NiTi wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

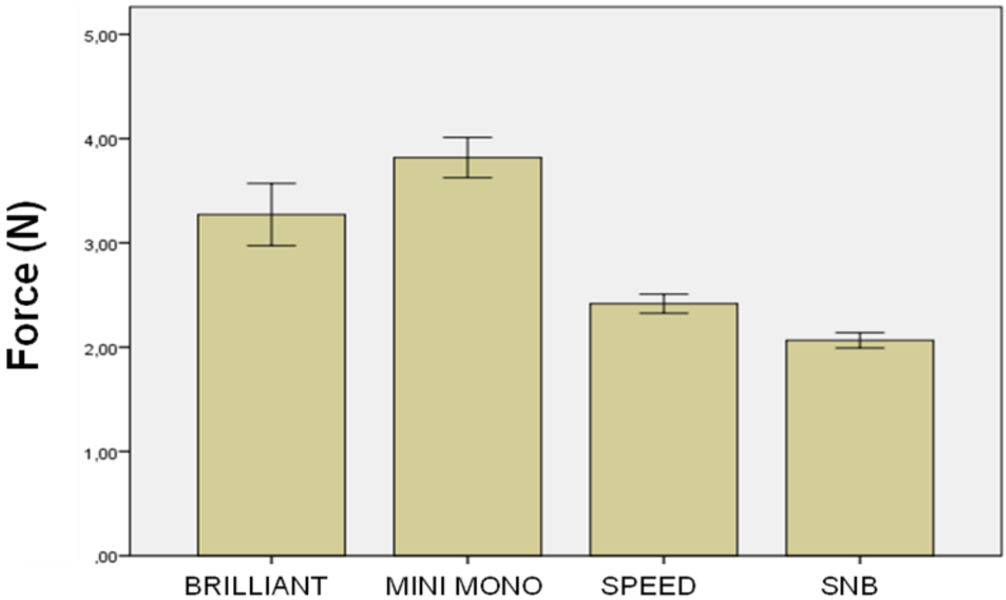

Fig. 25: Maximal forces generated at the beginning of the alignment and leveling phase in the x-axis by application of a 0.0135" NiTi wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

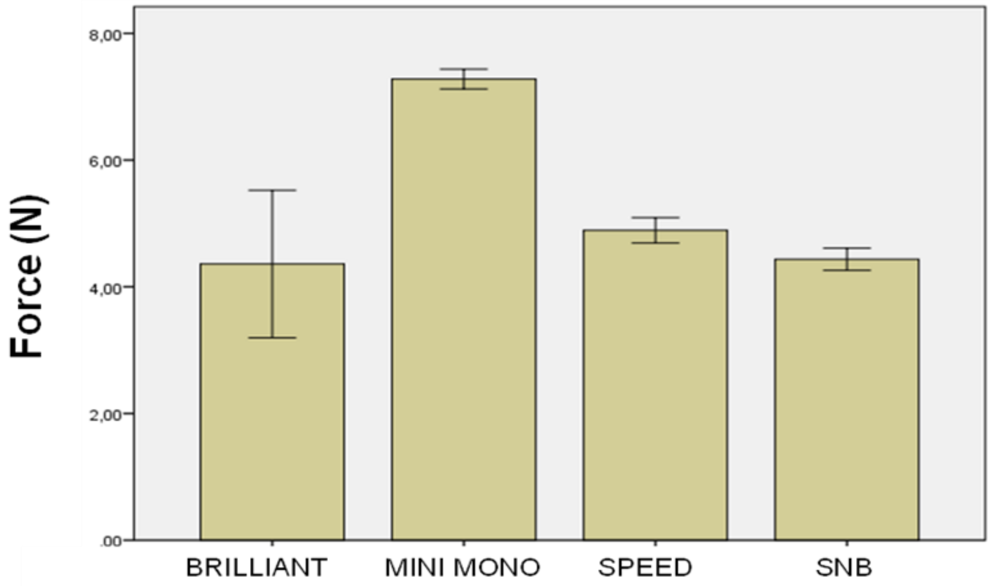

Fig. 26: Maximal forces generated at the beginning of the alignment and leveling phase in the x-axis by application of a 0.016" NiTi / Tröster Applications wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

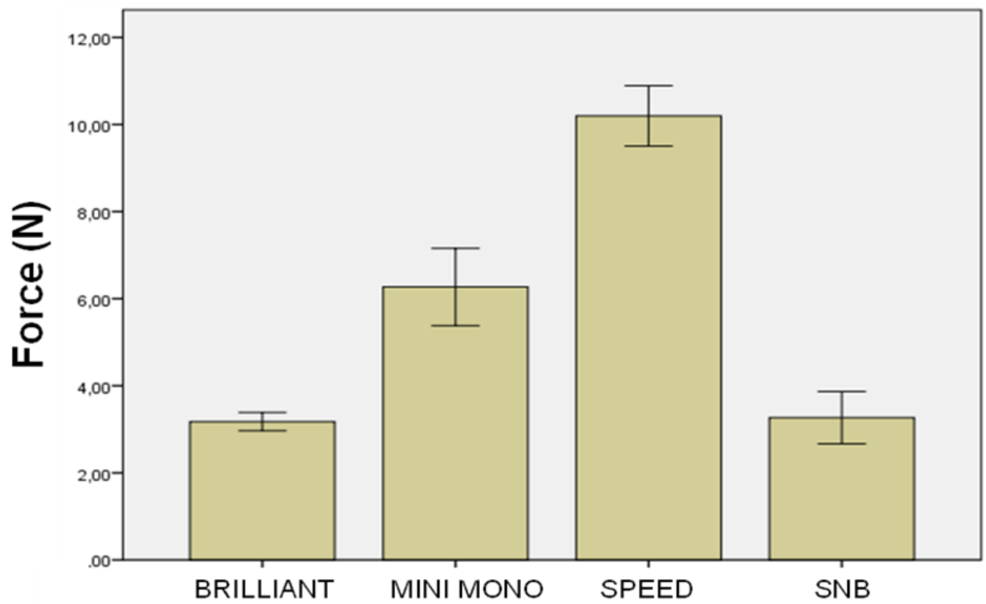

Fig. 27: Maximal forces generated at the beginning of the alignment and leveling phase in the x-axis by application of a 0.016" NiTi / BioStarter wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

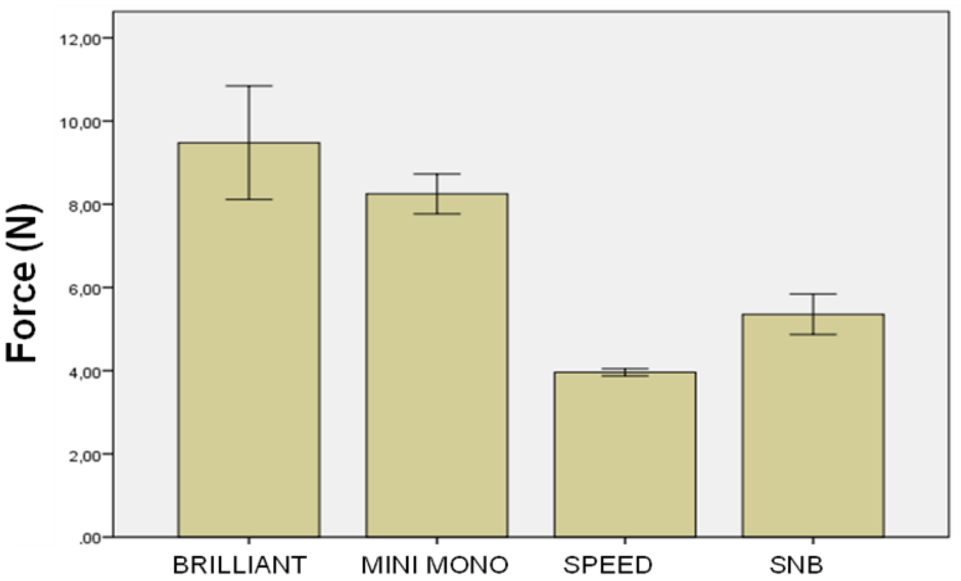

Fig. 28: Maximal forces generated at the beginning of the alignment and leveling phase in the x-axis by application of a 0.016" stainless steel wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

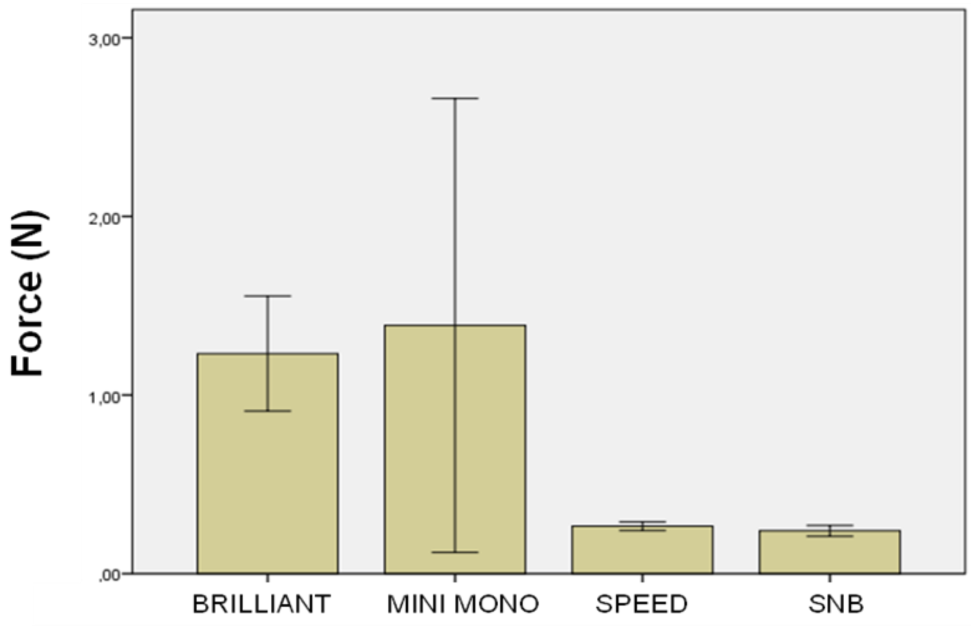

Fig. 29: Maximal forces generated at the beginning of the alignment and leveling phase in the z-axis by application of a 0.007" NiTi wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

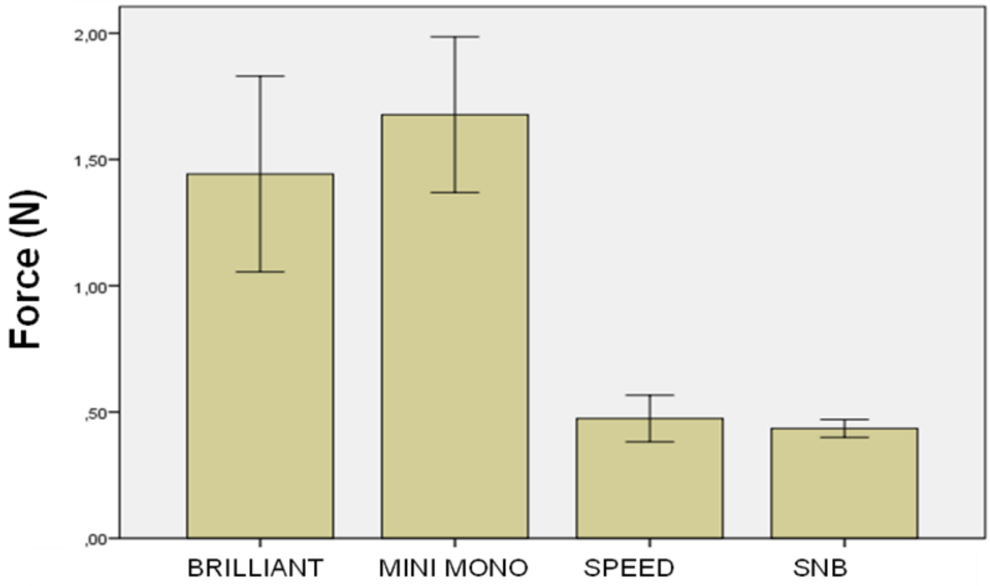

Fig. 30: Maximal forces generated at the beginning of the alignment and leveling phase in the z-axis by application of a 0.009" NiTi wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

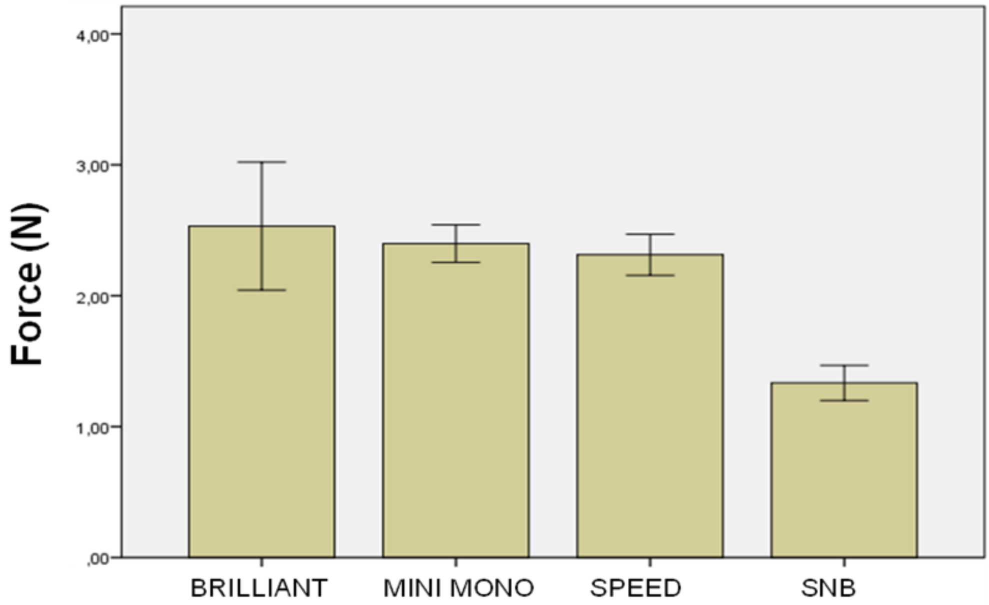

Fig. 31: Maximal forces generated at the beginning of the alignment and leveling phase in the z-axis by application of a 0.0135" NiTi wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

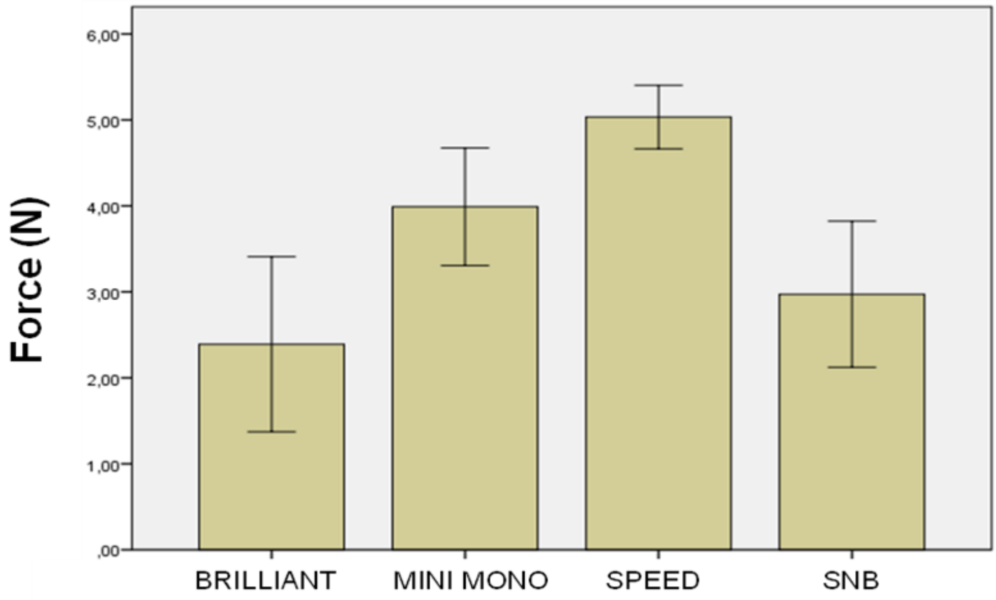

Fig. 32: Maximal forces generated at the beginning of the alignment and leveling phase in the z-axis by application of a 0.016" NiTi / Tröster Applications wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

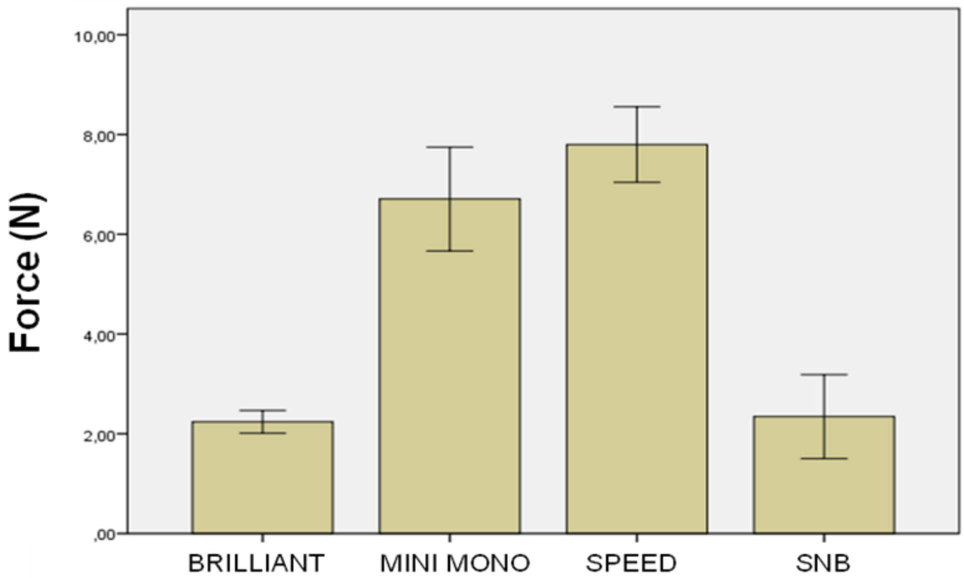

Fig. 33: Maximal forces generated at the beginning of the alignment and leveling phase in the z-axis by application of a 0.016" stainless steel wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

In the x-axis the generated forces range from 0.2 N up to 10 N. The lowest forces are expressed by using self-ligating brackets combined with small archwire cross sections. The insertion of 0.007" and 0.009" NiTi in the slot of a self-ligating bracket results in magnitudes of maximal 0.3 N. The same measurements concerning the conventionally ligating brackets are multiply higher and reach values up to 1.2 N. Furthermore, thicker archwires, which engage almost the whole slot surfaces, express forces of greater magnitudes. For instance, by using 0.0135" NiTi archwires the forces vary from 2.2 N (self-ligating brackets) and can reach up to 4 N, as far as the conventionally ligating brackets are concerned. A gradual increase of force magnitudes along with insertion of archwires with greater cross sections is observed. This finding precludes the usage of 0.016" NiTi alloy, which displays an extraordinary increase of forces on the teeth by the combination with SPEED brackets. Moreover, the comparative assessment of the provided data notes that the qualitative behavior of the force magnitude in the x- and the z-axis display many similarities, whereas by a quantitative comparison the forces developed in the inciso-gingival direction are higher as the one exerted in the labio-oral direction.

The above main observations concerning the values of initial maximal forces generated during the primer stages of an orthodontic tooth movement can be easily summarized in the two following diagrams (Fig. 34 a, b).

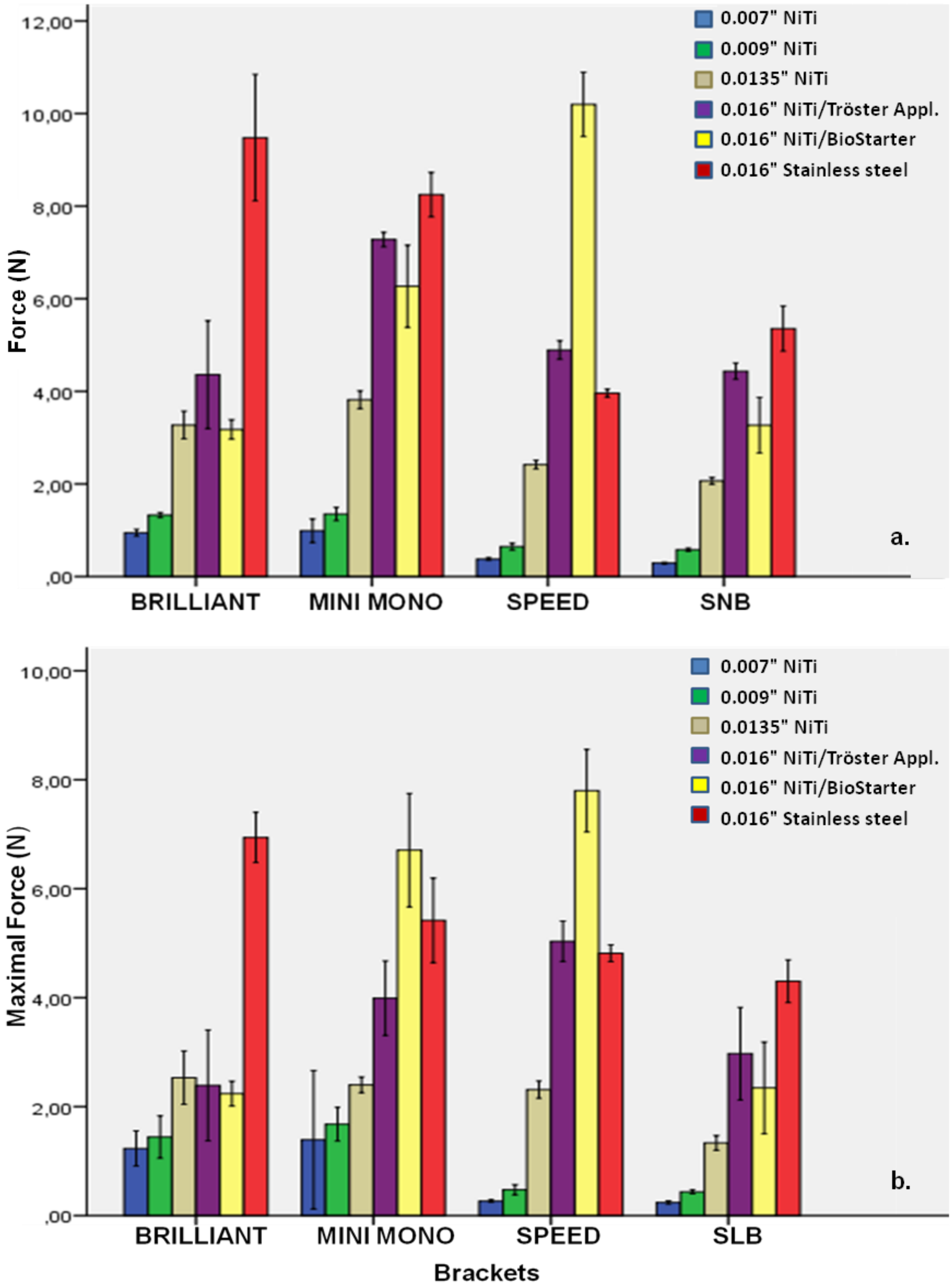

Fig. 34: Pool diagram of the maximal forces generated in the inciso-gingival direction (a) and in the labio-oral direction (b) on a maxillary incisor initially during a mock orthodontic tooth movement by combining each bracket design with each one of the investigated archwires. Standard deviation is also integrated in the graph.

5.2 Torque capabilities

Each experiment has been fulfilled after about 80 measurement cycles. In each one of these measurement cycles the moments generated after the application of a rotational movement of 20 degrees along the central axis of the slot have been recorded by means of OMSS in increments of 0.025 degrees. In the following diagram (Fig. 35) the process of torque activation will be presented indicatively in the y-axis. Through these curves the torque capability (maximal moment values) of the specific archwire, as well as the torque play of the specific wire engaged in the slot of a SNB were evaluated.

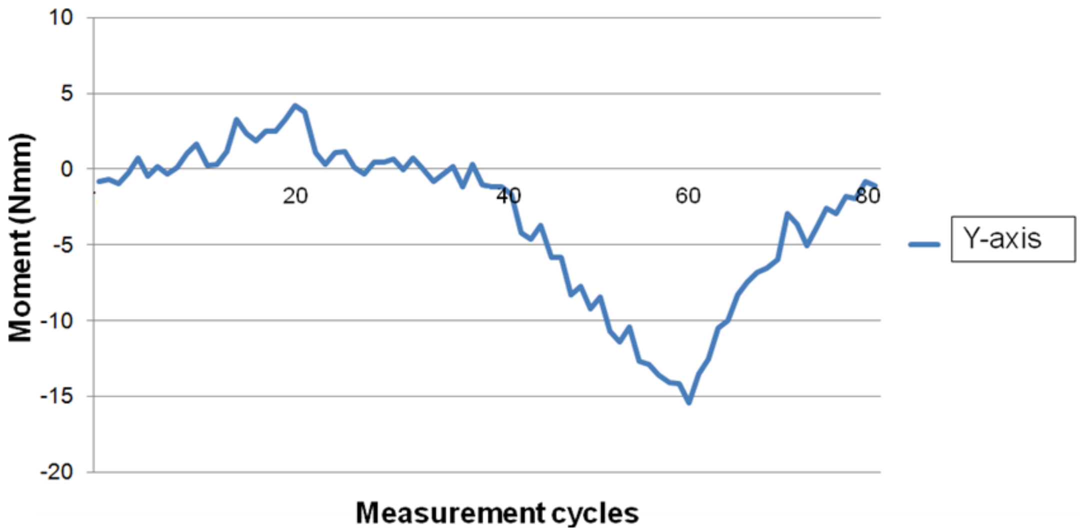

Fig. 35: Moments curve of a tooth under the application of a 0.016" NiTi / Tröster Appl. in the slot of a SNB in terms of a complete measurement with the OMSS.

5.2.1 Generated moments

In the following graphics moments (Fig. 36 – 39) generated after application of a rotational movement of ± 20 degrees on the maxillary incisor are presented analytically.

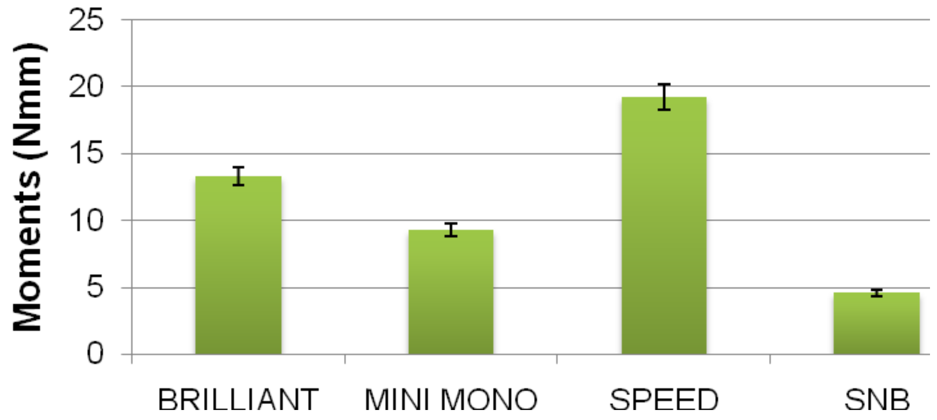

Fig. 36: Maximal moments generated during torsional movement under the application of a 0.016" x 0.016" stainless steel wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

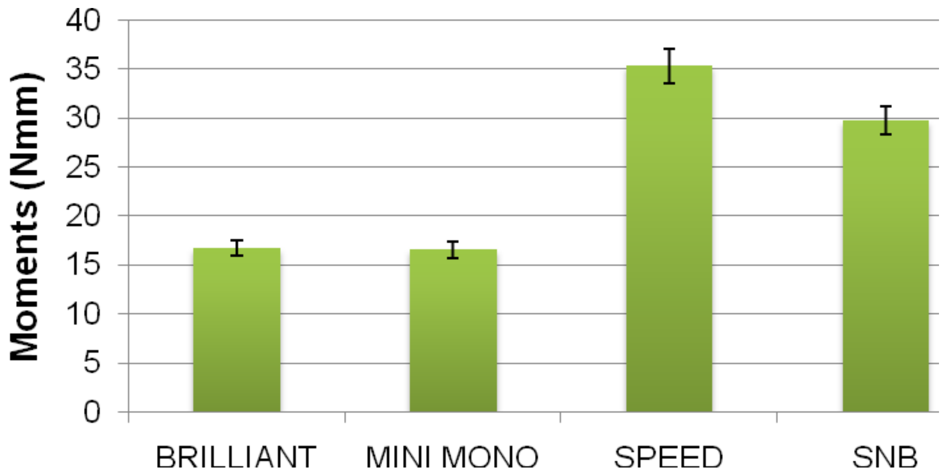

Fig. 37: Maximal moments generated during torsional movement under the application of a 0.016" x 0.016" NiTi/BioTorque wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

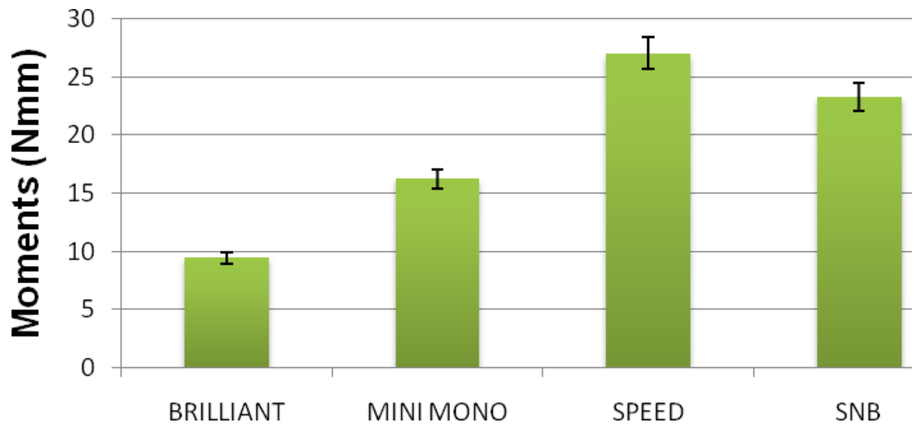

Fig. 38: Maximal moments generated during torsional movement under the application of a 0.016" x 0.022" stainless steel wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

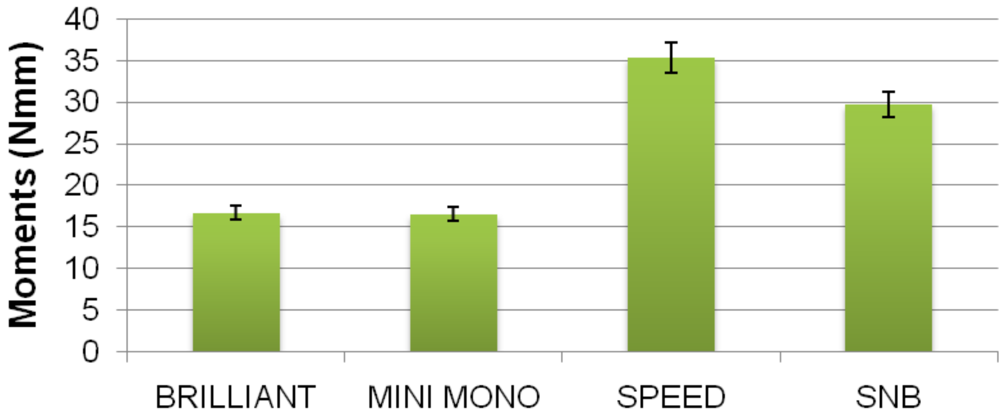

Fig. 39: Maximal moments generated during torsional movement under the application of a 0.016" x 0.022" NiTi/BioTorque wire in the slot of each of the investigated brackets. Standard deviations have been also integrated in the graphs.

According to the already provided data, the moments generated through the rotational movement of a maxillary central incisor range from 4.7 N (referring to SNB) and have reached values up to 35.2 N (referring to SPEED).

Generally, conventionally ligating brackets demonstrate lower torque capabilities in comparison to the self-ligating brackets and in particular to SPEED. The passive self ligating

SNB manifest similar torque effectiveness to the active self-ligating SPEED, precluding the engagement of a 0.016" x 0.016" BioTorque wire in its slot, where the values lie obviously in a lower level (4.7 N).

Presenting the obtained results on whole, a definite dependence of the torque expression is displayed on the cross wire dimension, rather than on the material. In fact, this finding corresponds all bracket designs. In almost all measurements, excepting for the 0.016" x 0.016" BioTorque combined with SNB, the results of the provided moment between the two different materials are quantitively very close, whereas a great diversity is demonstrated when an archwire with greater dimension is engaged in the corresponding slot.

The above main observations concerning the values of the generated moments during torsional movement, can be summarized in the following diagram (Fig. 40).

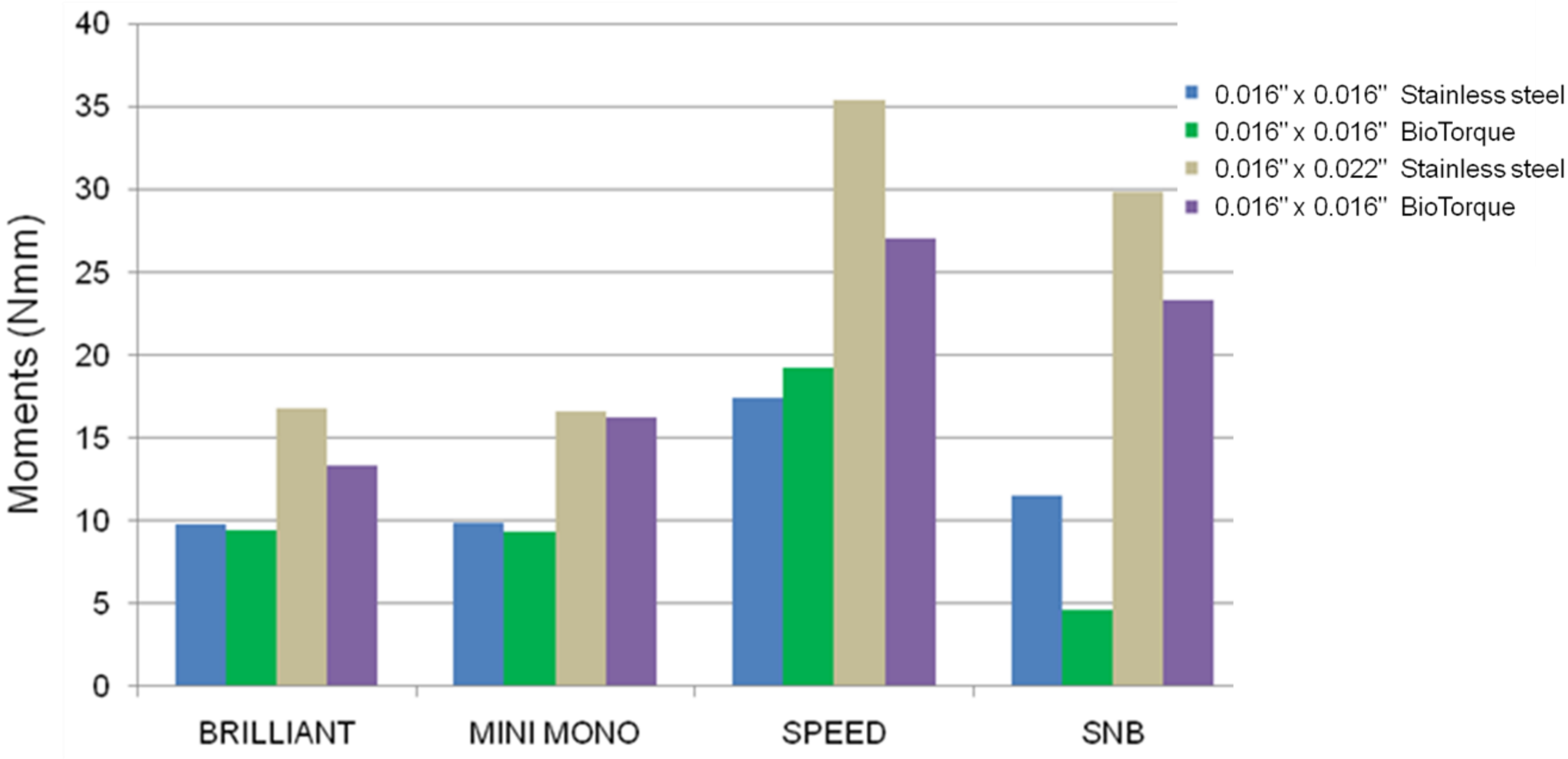

Fig. 40: Pool graph of the generated moments by combining all various bracket designs with all investigated archwires

5.2.2 Torque play

The above presented results referring to torque capabilities of various bracket systems can be correlated with findings related to the torque play. The graphic below presents in a comparative manner all the torque play values expressed by means of diverse bracket/archwire complexes (Fig. 41).

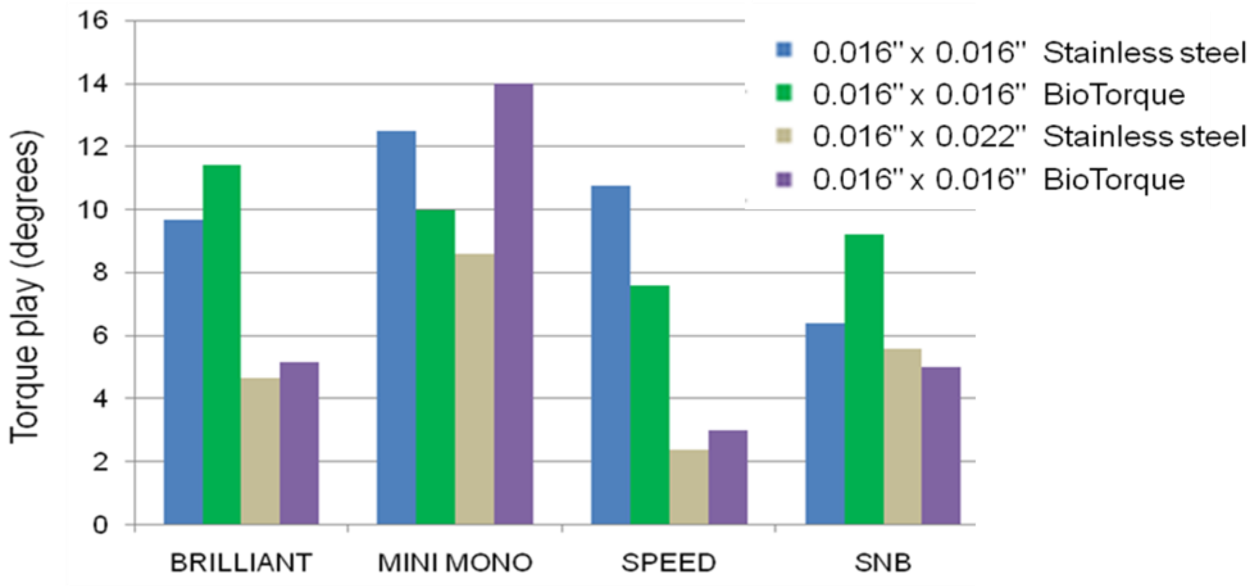

Fig. 41: Pool graph of the torque play by combining all various bracket designs with all investigated archwires

A first statement could underline the fact that the results are totally in alignment with the experimental outcome concerning the generated moments, which was previously presented. Torque deficiencies of the conventionally ligating brackets can be explained through the high levels of torque play. As far as the 0.016" x 0.016" cross section is concerned, the torque play fluctuates between 9.7 N and 12.8 N for the conventional brackets. For the self-ligating brackets the same measurements manifest lower values (6.2 N up to 10.4 N). However, as has been previously demonstrated, archwire which engage better the bracket slot displays subsequently lower torque play values. For instance, by the utilization of 0.016" x 0.022" NiTi wires, the torque play has been restricted in levels up to 8.5 N for the conventionally ligating brackets and 5.8 N for the self-ligating attachments. The only exception is recorded by the combination with the 0.016" x 0.022" BioTorque.

6. Discussion

In this chapter all the above presented results will be discussed and interpreted by means of a comparative assessment of the biomechanics of self-ligating appliances with the conventional systems, as is provided and evaluated in the current orthodontic literature. For a more comprehensive presentation of the factors affecting the obtained results the discussion will be divided into the following parts:

- methodology
- alignment and leveling effectiveness
- alignment and leveling forces
- torsional moments
- statistics

6.1 Methodology

The methodology of the current investigation included the construction of four identical resin replicas from an aligned mandibular model of a patient, in order to utilize them for the simulation of an orthodontic tooth movement. OMSS allowed us in each measurement cycle simultaneously, a three-dimensional registration of the force and moment systems, which affected the left mandibular incisor during the performance of a mock orthodontic tooth movement. OMSS facilitated also, the analysis of the correction of a complex malalignment in all three planes, observing at the same time the side effects which took place.

In contradiction to previously performed studies (Schumacher, 1992; Höse, 2007), where the investigated tooth movement concerned a unique-plane malposition, the current experimental set-up includes the assessment of force and moment levels as a result of a complex displacement. In particular, the specific range of displacement is confined to 2 mm in infraocclusion and 2 mm in an oro-vestibular direction, which approaches more realistic a clinical condition.

In-vitro studies, by default, present specific constrictions, which can potentially influence in a significant grade the clinical relevance of their results. As far as the current experimental investigation is concerned, the simulation of the periodontium and the function of teeth supporting soft-tissues could not be integrated in the performed measurements. Moreover, there has been set a fixed center of resistance for the maxillary incisor, as is provided in the study of Pederson et al. (1990). Nevertheless, this position can be variable in each patient and can be also related to the size of tooth. Furthermore, distribution of forces by interdental contact (deforming the alveolar ridge in conjunction with a reactive response from adjacent teeth) could not be taken into account in this in-vitro study.

This simulation of orthodontic treatment can be an indicative representation of what happens during actual orthodontic treatment, though precluding intraoral interactions. Hence, this study displays a model for the clinical situation, and our results describe a general trend of what is clinically observable. Assuming that these experimental tests cannot encompass all conditions in the clinical environment, nevertheless, this procedure provides, rather a closer approximation of realistic conditions than purely-static studies do (Schumacher et al., 1992).

6.2 Alignment and leveling effectiveness

The main goal of the first phase of orthodontic treatment can be summarized in the correction of slot differences, in order to achieve sagittally, as well as vertically, an ideal tooth arch. In course of the current study, the impact imposed by the different bracket systems combined with various archwires on the alignment and leveling effectiveness has been investigated.

Analyzing the obtained results on the whole, it is obvious that the various ligation types applied in each bracket system exert a significant influence on the degree of the malalignment's correction. Specifically, the degree of correction concerning the self-ligating brackets ranges from 72 % - 83 % and 76.7 – 98 % in the x- and the z-axis, respectively, whereas the same measurements for the conventional brackets demonstrate values of 70.9 – 77.39 % and 51.7 % - 58.5 % in the x- and z-axis respectively. The constricting alignment capabilities by the conventional brackets can be attributed to the frictional forces generated between bracket and archwire.

Friction is generally affected by the physical characteristics of the archwire and bracket materials, bracket design, archwire dimensions, as well as the method of attachment between archwire and bracket (Michelberger et al., 2000; Pizzoni et al., 1998; Schumacher et al., 1999; Thorstenson et al., 2001 Tidy et al., 1989). Additionally, elastic ligatures when are combined with conventional brackets can impose a positive effect on the expression of frictional resistance (Khambay et al., 2004). A single elastic module produces a ligation force of 50 to 150 g (Sims et al. 1993). The ligation method affects frictional resistance and is proportional to the force of ligation and the coefficient of friction of the contacting surfaces (Thorstenson et al., 2002). The relative magnitude of friction of ligation might, thus, vary according to the clinical situation (Tidy, 1989). The interaction between type of ligation and subsequent friction was the main objective of the study of Schwartz (2002), who concluded that the ligation method plays a more significant role in the generation of friction as the archwire dimension itself.

In order to restrict the problem of frictional resistance developed by elastic modules, various methods, including self-ligating brackets, have been suggested (Henao et al., 2004; Khambay et al., 2004; Thorstenson et al., 2001). In alignment with this point of view, the minimal inciso-gingival correction brackets observed in our study with the use of self-ligating brackets was 69.76 % and the lowest labio-lingual correction was 69.35 %. The only exception of these findings concerns the SPEED brackets combined with a 0.016'' stainless steel archwire, where the provided labio-lingual correction was only 11 %. This finding cannot be indicative for the general behaviour of self-ligating brackets regarding their frictional properties. However, it can be attributed to the aging of the locking mechanism, as the measurements concerning the 0.016 stainless steel archwire were the last to be performed.

The aging of the nickel-titanium clips depends significantly on the alloy composition and may produce a subsequent alteration of stiffness in a scale of 50 %. The clinical relevance of these findings may relate to the inability of the relaxed clip to apply forces due to aging, which may have been imposed by the mechanical loading or the environmental conditions, in general (Pandis et al., 2007).

Furthermore, comparing on the whole the two directions of the malalignment, better results can be noted in conjunction with vestibular displacement correction, than with correction of an infraocclusion. In fact, over-correction of vestibular displacement was apparent only with the SNB, by insertion in their slot of a 0.009" NiTi and 0.0135" NiTi. A unique case of overcorrection was referred by SPEED brackets combined with a 0.0135" NiTi inserted in both slots. In all these cases the central incisor was eventually not in the desired position in the dental arch, but rather slightly retruded. Overcorrection has been also observed in the x-axis expressed by the SNB by engagement of 0.0135" NiTi and 0.016" NiTi / Bio-Starter archwires in both slots, as well as 0.016" stainless steel wire in a single slot. This finding contradicts the results of the experimental study of Fansa (2009), which had presented overcorrection only in the oro-vestibular axis and no one in the vertical plane.

Nevertheless, the fact of the overcorrection can be correlated with the generated torque movement, which is manifested as a side effect during the orthodontic tooth movement. According to the Fig. 22 SNB manifest a consistent rotational movement between 5 ^{o} and 7 ^{o} combined with all archwires (with an exception of the double engagement of the slot with a 0.0135" NiTi). Similar findings are referring to the SPEED brackets display a torque movement of up to 8 ^{o}. From this magnitude’s spectrum insertion of 0.007" NiTi, 0.009" NiTi and 0.016" stainless steel wires is excluded. As far as the conventional brackets are concerned, the generated torsional movement displays lower values. The only exception is the Brilliant bracket combined with a 0.016" stainless steel archwire, which expresses an extreme rotation of 16 ^{o}, which leads to an oro-vestibular correction of 95 %. Therefore, it should be stated that the overcorrection is an indicator of a torque movement, which allows additional translation of the incisor’s edge, rather than an excessive levelling capability.

Moreover, the greatest degree of correction which has been noted in the labiolingual direction can be interpreted in terms of plastic deformation of the archwire, vertical play of the archwire in the slot or/and subliminal force magnitude through specific combinations of bracket/archwire.

In terms of the current study and in an attempt to investigate, whether simultaneous engagement of the main and the auxillary slot could ameliorate the control of the generated torsional movement, additional measurements were performed. Both the investigated self-ligating brackets have integrated a double slot, which were both engaged either with two 0.0135" NiTi, or 0.016" NiTi / Tröster Appl. or 0.016" NiTi / BioStarter wires in each measurement. According to the obtained results, the SPEED brackets displayed in both axes almost identical levels of correction, with the only exception the 0.0135" NiTi wire, which manifested slightly better alignment capability compared to other self-ligating brackets. As far as the torsional movement, as side effect, is concerned, no significant difference were recorded between engagement of a unique slot and both of them. The results which refer to the same measurements are differentiated for the SNB brackets, which demonstrated in both directions high levels of correction and sometimes overcorrection. The data regarding the torsional movement support the null-hypothesis and indicate lower values. Despite the fact that in the current study the passive self-ligating brackets, when both slots are occupied, displays obviously better biomechanical properties in comparison to the active ones, further clinical investigation is demanded in order to verify the suggested point of view.

From the aspect of the biomechanical behavior of archwires, the higher values in the x-axis are expressed by using an 0.0135" NiTi and an 0.016" NiTi / Tröster appl. archwire (96.97 % and 95.92 %, respectively). In the z-axis the use of 0.016" NiTi and 0.016" stainless steel resulted in the highest correction level (91.34 % and 98.8 %, respectively). The obtained results can be interpreted through the fact that extremely thin archwires display a very high value of play between the slot and the engaged wire, so that a great value of the exerted forces is lost and the bracket/archwire complex cannot express the maximum of its alignment capabilities. However superelastic archwires which display low values of frictional forces can be more effective in the correction of the same malalignment.

Regarding the descriptive statistics of the current study concerning the effect of the different bracket designs and types of archwires on the subsequent alignment and leveling effectiveness, a significant effect of both factors is displayed. The provided results also indicated the presence of interaction ($p \leq 0.000$) between the two variables. Therefore, according to the current study, utilization of passive self-ligating brackets combined with NiTi alloys of relatively small cross sections would be preferable in order to maximize the effectiveness of the correction of a malalignment. The current suggestions converge partially on the conclusions reached in the studies of Andreasen et al. (1980), Kusy et al. (1987) and

West et al. (1995), who observed significant interaction between archwire and enhanced alignment and leveling capabilities, too.

6.3 Alignment and leveling forces

The orthodontic forces commonly proposed for tipping and extrusive movements range from 0.35 to 0.6 N (approximately 35 – 60 g). The lower value is recommended for incisors and the higher ones for multirooted teeth (Luppanapornlarp, 2010; Owman-Moll, 1996; Proffit, 2007; Stonner, 1960). The forces recorded in the current experimental investigation both in the x- and the z-axis, representing an extrusive and a labial movement respectively, were generally much higher as the recommended ones. In a revision of the literature comparable measurements performed likewise with the OMSS demonstrate high initial force magnitudes during the correction of a complex malposition (Fansa, 2009; Holtmann, 2012; Montasser, 2013) ranging from 0.6 up to 8.1 N.

In the current study the values referring to the mean maximal forces in the inciso-gingival axis by using the 0.007" NiTi and the 0.009" NiTi combined with all bracket designs did not overcome 0.88 N. However, the relevant values by utilizing the 0.0135" NiTi archwires, as well as those with wider cross sections, were significantly higher and reached magnitudes up to 6.55 N.

As far as the z-axis is concerned the results were similar with the inciso-gingival axis, though the values slightly lower, as the achieved maximal force was evaluated to be 5.21 N. The degradation of the force values as is presented in this study is in alignment with the view that wire stiffness results in higher forces (Drenker, 1998).

However, through the results of this experimental investigation, it should be clearly stated that the ligating mode prevails over other affecting factors. SNB exert the lowest forces with a magnitude of 2.42 and 1.76 N in the inciso-gingival and labio-oral direction respectively, whereas the Mini Mono the highest ones, with the corresponding values approaching a magnitude of 5.19 and 3.71 N in the x- and z-axis, respectively. The drop in the forces exerted by the self-ligating brackets may be assigned to the increased play of wires in the slot and the lack of obstacles arising from the contact of an elastomeric ligature out-side the wings. The difference accounts for about 2 N or about 100 % of the forces observed with self-ligating brackets and therefore the superiority of the self-ligating brackets against conventional ones could be clearly expressed.

Additionally, a pivotal role in the interpretation of the obtained results plays the friction developed between the archwire and the slot. It is also of great importance the correlation of the forces generated with the consequent degree of correction, taking into consideration the generated friction. For instance, in the x-axis by forces of around 2 N with insertion of 0.007" or 0.009" NiTi archwire in the slot of conventional brackets the provided correction ranges between 15 and 58 %. On the contrary, the same measurements with the utilization of self-ligating brackets present a 50 % reduction of forces (maximum 1.1 N) with a consequent correction always over 60 %. These findings are in alignment with the studies of Berger (1990) and Matarese et al. (2008), who, despite using different experimental measurement system as the one used in the current investigation, supported also the point of view that the magnitude of forces by using self-ligating brackets in comparison to conventional ones is significantly lower.

Nonetheless, friction is regarded to be only a part, and usually a small one, of the resistance to movement as a bracket slides along an archwire. According to Kusy et al. (1997) and Pizzoni et al. (1998) resistance to sliding (RS) could be also a result of a permanent deformation of the wire, which occurs at the wire bracket corner interface, and is defined as notching. This phenomenon has also been in this study observed, as at the end of the alignment and leveling process and having ligating an archwire in the slot of conventional brackets with elastic ligatures, a plastic deformation took place resulting in a high magnitude of forces and restricting degree of correction of the malposition.

Furthermore, the results of our study imply that interbracket distance could be a predicting factor of force magnitude during archwire engagement. This is clearly illustrated in the results found for the bracket with the least width (Swiss Nonligating Bracket), which showed force levels significantly lower than those by Mini Mono brackets, who display the maximal width among the investigated orthodontic attachments.

Another interesting finding is the difference of the generated forces by the engagement in the various bracket slots of two 0.016" NiTi archwires, provided from two different producers. This phenomenon could be explained through the development of diverse frictional forces between the bracket and the archwire in interaction to their biochemical properties. According to Brady (1995), adding copper to the alloy lowers the friction of the NiTi wire and makes it slip more easily along a bracket. On the contrary, an investigation of Kusy et al. (1997) has suggested that the Cu-NiTi can generate higher friction than the A-NiTi wire, in terms of differences in surface chemistry and chemical affinity between these two types of archwires. In addition to that, the bending stiffness of the archwire is considered to be one of the parameters that also correlates with generated forces. Bending stiffness is directly associated with the nominal dimension of each wire (Henao et al., 2004). However, further studies are needed to shed light on the expression of this phenomenon.

Conclusively, can be clearly stated that by means of OMSS a clinically realistic assessment of the acting forces during the procedure of correction of a malalignment can be presented. However, it should be underlined that the provided values cannot be fully transferred in the clinical practice, but should be though used as a reference of a qualitative and not quantitative comparison of the effectiveness of the diverse bracket / archwire.combinations.

6.4 Torsional moments

Correction of axial variation of teeth requires a moment applied to the bracket to initiate a rotational movement. Factors that can result in torque variations include manufacturing process effects, varying material properties, and clinical procedures including ligation methods and interbracket distance (Huang et al., 2009).

According to the available literature (Bantleon et al., 1988; Burstone, 1966; Feldner et al., 1994) effective values for rotational movements ranging between 10 and 20 Nmm. Currently, there is a lack of evidence on the torque characteristics of various bracket/archwire combinations (Alkire et al., 1997; Harzer et al., 2004). This may be attributed to the complexity of the experimental configuration required in laboratory studies, and the multiplicity of factors needed to be controlled in a clinical setting, including individual response to moments applied, variability in malocclusion and the potential effect of other auxiliaries or treatment utilities in affecting torque (Huang et al., 2009).

The moments generated in the current study approach principally the limits set from the previous investigations, including though some restrictions. In particular, all the evaluated moments exert greater magnitude than 10 N with an exception of the SNB combined with 0.016 x 0.016" BioTorque archwire, where the values are only 4.9 Nmm and cannot display sufficient clinical efficiency. On the contrary, SPEED brackets demonstrate enhanced torque capabilities by expressing torsional movement of 17 Nmm by application of a 0.016" x 0,016" stainless steel archwire and have reached the value of 35.2 Nmm by ligation of a 0.016" x 0.022" stainless steel archwire.

According to a previous study (Huang et al., 2009), it was indicated that ligation methods of the various bracket designs can impose an impact factor on the torquing moment. On the one hand, it has been found that self-ligating brackets compared with conventional ones displayed higher torque play and, thus, reduced torque capability. On the other hand, self-ligating and conventional brackets have different widths, which in turn is a contributing factor to the torque capabilities.

In addition, as has been previously referred, the generated moments are mainly influenced through the stiffness of the inserted archwire rather than its material. The change from 0.016" x 0.016" to 0.016" x 0.022" archwire increased predominantly the torque moments, rather than the change of the wire property from nickel titanium to stainless steel. The torsional stiffness of the arch wire is the significant contributing factor with regard to the torquing moment. Nevertheless, the narrow size of the self-ligating brackets compared to the one of conventional brackets attributes a better torque capability to the self-ligating brackets. This can be explained due to the fact that the bracket width influences the free wire length. Although the torsional stiffness of an archwire -and thus the moment/torque characteristic- is dominated by the wire cross section, altering the free wire length could be another method proposed to control torque expression. This aspect coincides with the finding of our study, where self-ligating brackets with a decreased bracket width increase the torquing moments compared to conventionally ligating brackets.

Apart from that, the results of the generated moments seem to be dominated by the wire slot/play, in general, as well as by the special effect of the active NiTi spring during torque control, in particular. The function of the NiTi spring can be summarized in the pressure exerted on the wire onto the bottom of the slot of the Speed bracket and its subsequent activation by the wire upon torquing. The contribution of the NiTi clip in the superior biomechanical behavior of SPEED brackets against SNB, can be also attributed to the function of the active clip. On the other hand, the rigidity of the closing component of the buccal slot wall may limits the available space for the wire to move and dissipate some of the energy given at engagement. This behaviour can be easily explained by the common deformation of the superelastic NiTi clip and the wires engaged into the bracket slot. The same occurs with conventional brackets owing to the deformation of the elastomeric ligatures, which show more relaxation.

The elastomeric ligatures are considered to be the reason of restricting torque capability results, as far as the conventionally ligating brackets are concerned. This concern is particularly relevant to elastomeric modules, since polyurethane-based elastomers have been found to lose generally approximately 50% of the force applied within the first 24 hours in an in vitro set-up (Taloumis et al., 1997). More decay is expected in the oral environment because of the severity of conditions existing in the presence of pH fluctuations, temperature variations, enzyme action and mechanical loading. Hence, their use as a ligating medium in rotational movement has been questioned and stainless steel ligatures have been suggested for more efficient and consistent engagement (Eliades et al.; 2009). However, conventional ligation offers the advantage of modulating the extent of ligation by using elastomers in figure-of-eight configurations, or using stainless steel ligatures with varying degrees of tie-in force. Nonetheless, the use of the latter is associated with the development of higher moments, which may exceed the biological range (Hemingway et al., 2001).

Moreover, aging alterations of the clip of passive self-ligating bracket occurring during the course of orthodontic treatment may modify the moments generated during wire engagement (Pandis et al., 2007).

As far as the torque play is concerned, the interaction between high levels of torque play and restricting values of generated torsional moments can be clearly stated in the course of this study. The greatest torque play was registered by the utilization of Brilliant and Mini Mono brackets (9.5^{o} - 14^{o}), where has been also demonstrated the lowest levels of moments (10 – 16 N). On the contrary, SPEED brackets, by manifesting the best torque capabilities in comparison to all other bracket designs, have consequently expressed a torque play between 2 ^{o} and 11 ^{o}; values significantly lower compared with those of the rest investigated brackets. In fact, most of the studies evaluate this biomechanical property of fixed orthodontic appliances by investigating the values on a single rotated bracket and the wire into account. However, a further factor which tends to lower the torque moment of a wire under torsion in a bracket/archwire system is the play of the wire within neighboring brackets. Nevertheless, this factor could not be incorporated in the course of the current study.

Additional clinical factors, which influence torquing moments and could not be studied in the current investigation, is the accuracy of vertical bracket positioning and the morphology of the teeth. In particular, according to Meyer et al. (1978) vertical shift of 3 mm can change the torque angle by around 15 degrees. Miethke (1997) on the other hand, proposed that a torque variation of 10 to 15 degrees may already stem from a vertical inaccurate placement of 1 mm. Moreover, the morphology of the teeth can vary greatly and consequently affect the clinical use of torque (Morrow et al., 1978).

6.5 Statistical analysis

The ANOVA results proved that the type of the bracket used had a significant impact ($p<.05$) on the alignment and leveling effectiveness, as well as on the magnitude of forces exerted in the initial phase of the correction of an orthodontic malalignment (Table 5). In particular, the deviations for each type of the various archwires combined with the four bracket designs were significant. There is no available pattern distinguishing the different results of the alignment capability between self-ligating and conventional brackets, as well as between active and passive self-ligating attachments.

Source	Sum of squares	Df	F	Sig. (p-value)
Brackets	**13886.271**	**3**	**77.849**	**0.000***
Wires	**26763.136**	**5**	**90.023**	**0.000***
Brackets x Wires	**18557.406**	**15**	**20.807**	**0.000***
				Significance < 0.05

Table 5: Results of the two-way ANOVA regarding the correction (%) of a malposition for the different bracket / archwire combinations.

The same type of analysis examined the correlation of both factors affecting the initial maximal forces generated during an orthodontic tooth movement.

According to the two way ANOVA the type of wire combined with the bracket had a significant effect on the maximal forces developed (Table 6). For each bracket among the 4 different ones included in the study, the mean maximal values were significantly different, when the bracket was used in combination with each of the various wires that were investigated.

Source	Sum of squares	Df	F	Sig. (p-value)
Brackets	**6.388**	**3**	**27.392**	**0.000***
Wires	**2.392**	**5**	**6.154**	**0.060***
Brackets x Wires	**15.247**	**15**	**13.076**	**0.000***
				Significance < 0.05

Table 6: Results of the two-way ANOVA regarding the maximal initial forces exerted on tooth for the different bracket / archwire combinations.

7. Conclusions

In the initial phase of alignment self-ligating brackets combined with archwires of a small cross section dimension display distinct advantages, as far as the degree of correction of a complex malposition is concerned. On the contrary, conventional brackets ligated with elastomeric modules present a number of side effects which can affect determinatively the outcome of the orthodontic movement.

The superiority of the self-ligating brackets against the conventional ones is suggested through this study, as far as the exertion of light and constant forces is concerned. The magnitude of forces is significantly lower in comparison to the one generated with attachment of conventional brackets, resulting in an orthodontic outcome which would not jeopardize a good prognosis of the supporting tissues.

In the current study on the whole, the active self-ligating brackets have demonstrated slightly better results as the passive ones. For instance during rotational movements SPEED brackets manifest a constricting degree of torque play and subsequently effective torque capability comparing it with the passive self-ligating brackets, as well as with the conventional ones. The qualities of the NiTi spring play a prominent role in the demonstration of these advantageous biomechanical properties. However, the grade of the statistical significance do not allow us to verify this point of view, and that is the reason that further investigation is needed in order to provide answers to this question.

Additionally, it should be underlined that the greater impact factor on the expression of better torque capabilities is imposed by the archwire dimension, rather than its material properties.

Potential differences in the values reported in this study and previous investigations examining self-ligating brackets should be assigned primarily to different types of closing mechanisms appearing among self-ligating appliances, different bracket widths and varying bracket slot–archwire play, as well as diverse methodology.

Despite the fact that self-ligating brackets are not regarded as new inventions of the last years, they remain a controversial topic amongst the scientific society. Moreover, in orthodontics the necessity of applying light forces has always been emphasized. However, there is a notable scarcity of evidence on the force and moments generated during activation of an archwire in self-ligating brackets in a crowded tooth arch. It can be easily understood, that it is of great importance to be able to describe the biomechanical ramifications and engineering limitations of each bracket design. Therefore, the aim of each clinician should be the application of effective appliances, which would not jeopardize in any way the clinical outcome, but would minimize the treatment's duration and ensure simultaneously the patient's convenience.

In order to confirm or refute this study, it would be therefore useful in the same context the force ratios, the grade of correction achieved, as well as the torque effectiveness to be part of an investigation, taking also in account physiological factors such as the periodontal ligaments, and interproximal contacts.

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