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Naseeba Khouja

**In Search of An Improved Methodology for Measuring Dental Implant  
Stability:  
Combining Experimental Model Analysis (EMA) and Finite Element Analysis  
(FEA)  
to Achieve Maximum Reliability**

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University of Washington

**ABSTRACT**

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Despite the importance of the need for measuring dental implant stability, an effective, reliable methodology has eluded researchers, with current methods used to detect implant stability either in the early stages or incapable of providing a complete picture. This study sought to identify a parameter capable of measuring dental implant stability, to allow for the successful detection of

failing implants. Through the combined use of experimental modal analysis (EMA) and finite element analysis (FEA), we examined the accuracy and reliability of three implant stability devices—the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter—by calibrating them via both EMA and FEA, using models of dental implants of various lengths in Sawbones<sup>®</sup> of different densities. The theoretical mechanics used to operate the three devices were analyzed to understand their assumptions and limitations. To estimate angular stiffness, we employed implant models of various lengths and widths in Sawbones<sup>®</sup> of different densities.

To gauge how effectively Sawbones<sup>®</sup> and porcine bones could represent human bones, different types of bone models (i.e., porcine bone, human mandibular bone, and Sawbones<sup>®</sup>) were compared using the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> devices, and EMA, in addition to measuring apparent bone density. Micro-computed tomography (micro-CT) was used to estimate bone mineral density (BMD) and morphometric parameters of both porcine and human bone samples. After demonstrating that the three above-mentioned devices were deficient both theoretically and experimentally, we developed an optimal model for implant stability testing through the combined use of EMA and FEA. Natural frequency, although a robust and consistent measurement, does not equate with dental implant stability. Porcine bone may overestimate implant stability measurements when applied to human bones. Sawbones<sup>®</sup> may provide reasonably good models but should be used with caution. Boundary conditions and experimental setup play an important role in the outcome of research and should be taken into consideration. The angular stiffness coefficient—which is independent of the type of abutment used—was found to represent a superior index for quantification of dental implant stability, successfully differentiating stability of implants of both varying lengths and diameters, while the other methods tested did not.

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## **CHAPTER 1. INTRODUCTION**

Dental implants are widely used for replacing missing teeth, and are considered the treatment of choice in cases where conservation of bone and teeth is desired (Buser et al., 2017; Haiat et al., 2014; Quirynen et al., 2014). During the past decade, the use of dental implants to replace missing teeth has increased dramatically. According to The American Academy of Implant Dentistry, it is estimated that 500,000 dental implant procedures are done each year, with approximately 3 million individuals in the US having benefited from this procedure to date (AAID). Dental implant therapy has become one of the most successful and predictable treatments performed in dentistry. However, as is the case for all treatments, failures of implant integration still occur and remain difficult to predict (de Freitas et al., 2012; Esposito et al., 2013; Esposito et al., 2007b; Lang et al., 2012; Mathieu et al., 2014; Vina-Almunia et al., 2012). The long-term maintenance and monitoring of dental implants has been an inexact clinical science. The failure of an implant by epithelial down growth and loss of osseointegration after years of successful function (Buser et al., 2017; Chrcanovic et al., 2014; Engelhardt et al., 2015; Figuero et al., 2014; Jensen, 2015; Koldslund et al., 2010; Quirynen et al., 2014; Romanos & Weitz, 2012; Temmerman

et al., 2014), or as a result of “micromotion” (Brunski, 1999; Brunski et al., 1979; Kheur et al., 2016), or because of excessive mechanical forces applied too early to the implant (Atsumi et al., 2007; Ericsson et al., 2000; Esposito et al., 2013; Esposito et al., 2007; Grutter & Belser, 2009; Hsu et al., 2013b; Ioannidou & Doufexi, 2005; Javed et al., 2013; Javed & Romanos, 2010; Lindeboom et al., 2006; Lorenzoni et al., 2003; Nkenke et al., 2005; Shibly et al., 2012; Siddiqui et al., 2008; Su et al., 2014; Tealdo et al., 2011; Tsirlis, 2005; Turkyilmaz et al., 2012; Vayron et al., 2013; Zafiropoulos et al., 2010; Zhou et al., 2009) is not well understood.

Clinicians can readily recognize a healthy implant and easily identify a failed implant. The challenge relates to that part of the spectrum that falls between these two extremes. With few clinical indicators, clinicians tend to be weak with regard to recognizing when an implant has started on the path of compromise or failure, until there is significant radiographic bone loss indicating that intervention is called for. At this time in clinical implant dentistry, the zone of the implant health spectrum leading to failure is poorly defined and needs investigation (Chrcanovic et al., 2014; de Freitas et al., 2012; Engelhardt et al., 2015; Esposito et al., 2007a; Esposito et al., 2013; Haiat et al., 2014; Ioannidou & Doufexi, 2005; Jensen, 2015).

Although many studies have been carried out to assess dental implant stability, no standard protocol is available for clinical use. Current biotechnologies have proved of little value in identifying compromised dental implants. Dental practitioners have traditionally relied upon percussion and instrument tests to evaluate dental implant stability. Radiographs are routinely utilized for evaluating conditions relating to the critical aspect of bone-to-implant interface. Together, these serve as the most commonly employed diagnostic tools for clinicians. For this reason, there is a great need to develop more sensitive, early diagnosis and detection

methodologies for compromised implants or implants on a path to failure. The main objective of this dissertation is to define a parameter capable of measuring dental implant stability, a more precise and reliable parameter for detecting failing dental implants than is currently in standard use. This endeavor naturally begins with determining the biomechanical factors affecting dental implant stability and identifying the advantages and drawbacks of the current dental implant stability measuring devices.

The 7 central hypotheses on which this study is based are as follows: (1) The devices currently used for measuring dental implant stability are not as accurate as claimed by their manufacturers and are not able to detect the biomechanical characteristics of dental implant stability. (2) Finite element analysis (FEA) predictions agree with experimental modal analysis (EMA) measurements. (3) Stability is positively correlated with implant length. (4) Dental implants embedded in high-density blocks are more stable than implants placed in those of lower density. (5) The way the experiment has been set up and its boundary conditions can have a significant effect on results. (6) Wide-platform dental implants placed in low-density blocks provide more stability than regular-platform dental implants. (7) Angular stiffness represents an appropriate measure for dental implant stability.

These hypotheses are tested in the chapters below in the following two ways:

1. Define the key biomechanical factors affecting primary dental implant stability in vitro, using different implant lengths, widths, and different bone models. Specifically, this is done by developing a synthetic experimental bone model to evaluate the accuracy and ability, so as to detect the mechanical characteristics of primary implant stability. We also measure bone density in-vitro (with animal bone) using micro-computed tomography (micro-CT) ( $\text{g/cm}^3$ ).

Morphometric measurements such as bone-to-implant contact (BIC) will be done using micro-CT. We will also examine the effect of implant length, width, and bone density on dental implant stability measurements. In addition, we will examine how abutments with different lengths impact dental implant stability measurements.

2. Determine and identify the most reliable method of measurement that is sensitive to the different key factors relating to primary dental implant stability. We will analyze the mechanical theories, physics and equations of the testing devices (i.e., Periotest<sup>®</sup> [Medizintechnik Gulden e. K. Modautal, Germany], Osstell ISQ<sup>®</sup> [Osstell, Gotenburg, Sweden] and Quantitative Percussion Diagnostics, aka Periometer [Newport Beach, CA]). This analysis will be done by comparing the results of experimental modal analysis (EMA) and finite element analysis (FEA) predictions, with the results obtained by the commercially available devices, to determine accuracy. In other words, we want to use EMA and FEA as a standard or reference, for purposes of evaluating the commercially available measuring devices. EMA is a standard vibration testing procedure used in mechanical engineering to simultaneously measure natural frequency, viscous damping factor and stiffness coefficient of a tested specimen. FEA is used to simulate the experimental model and predict EMA results. FEA has the advantage of simulating various experimental conditions. Mechanical parameters (such as Young's modulus of elasticity) for each material must be entered carefully, as they will affect results. There is reason to suspect that there are number of fundamental issues with implant stability measuring devices currently in use (Aparicio et al., 2006; Atsumi et al., 2007; Haiat et al., 2014; Manzano-Moreno et al., 2015; Meredith et al., 1998; Pattijn et al., 2006). For this reason, we want to identify a more precise, reliable method of detecting the key factors relating to implant stability. We also want to confirm how natural frequency, stiffness and



damping ratio relate to dental implant stability. The resulting data will be used to generate a conceptual design for a device based on sound scientific and mechanical theories.

In the chapters that follow, we will:

- Define dental implant stability, types, and factors affecting stability. We will also examine the methods (invasive and non-invasive) currently described in the literature for measuring dental stability, pointing out the problems present in various published studies.
- Evaluate the accuracy and reliability of the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter, by calibrating them via both experimental modal analysis (EMA) and finite element analysis (FEA). In particular, measurements by the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter are compared with those obtained by means of both EMA and FEA. We will also analyze the theoretical mechanics used to operate the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter. Rigorous analysis allows us to understand assumptions and limitations of these devices. It also provides us with important clues to how to achieve implant stability from these measurements.
- Observe how well Sawbones<sup>®</sup> and porcine bones represent human mandibular bones. This is achieved by comparing the various types of bone models (human, animal, and Sawbones<sup>®</sup>), by measuring with the Osstell<sup>®</sup> ISQ, the Periotest<sup>®</sup> PTV, natural frequency values, and apparent bone density.
- Examine bone density and implant-bone morphometrics of the different types of bone models (human, animal, and Sawbones<sup>®</sup>), using  $\mu$ CT.

- Determine how boundary conditions and experimental set-up play an important role in dental implant stability testing.
- Describe a novel yet promising method for assessing the potential for dental implant stability
- Offer a summary of conclusions, as well as some ideas for future research.

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## **CHAPTER 2. BACKGROUND AND LITERATURE REVIEW**

In this chapter, we will define implant stability, types of dental implant stability, and factors affecting dental implant stability and osseointegration. We will also examine the methods used to measure dental implant stability. In addition, we will describe the problems described in the literature regarding implant stability testing.

### **2.1 What is Dental Implant Stability?**

Dental implant stability has been described as the absence of clinical mobility of an implant (Brunski, 1992; Cochran et al., 1998). Haiat et al. (2014) defined implant stability as “the capacity of an implant to carry loads in the axial and lateral directions as well as in rotation.” (p.189). From a biomechanical standpoint, implant stability results from the interaction between an implant surface and bone, creating the bone-to-implant interface.

## 2.2 Types of Dental Implant Stability

There are two different types of stability explained in the literature: primary (mechanical) and secondary (biological). These mechanisms are responsible for anchoring the implants in alveolar bone.

### 2.2.1 *Primary stability.*

Primary stability occurs as soon as an implant is inserted in bone. Such stability is mainly biomechanical in nature (Haiat et al., 2014; Javed et al., 2013; Marquezan et al., 2012; Mathieu et al., 2014). Primary stability depends greatly on the quality and quantity of bone at the host site (Bilhan et al., 2015; Cavallaro et al., 2009; Cehreli et al., 2009; Hong et al., 2012; Hsu et al., 2013a; Kang et al., 2016; Marquezan et al., 2012; Mohlhenrich et al., 2015; Romanos et al., 2016; Tolstunov, 2007; Torsiglieri et al., 2015; Turkyilmaz et al., 2009; Wang et al., 2016; Wang et al., 2015), implant design (e.g., implant shape, length, diameter, and thread design) (Bilhan et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gómez-Polo et al., 2016; Hong et al., 2012; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016; Sennerby et al., 2015; Tabassum et al., 2010; Wang et al., 2015; Yamaguchi et al., 2015), surgical technique used, and seating of an implant in bone (Ahn et al., 2012; Chan et al., 2010a; Degidi et al., 2015; Degidi et al., 2017; Sennerby et al., 2015; Sierra-Rebolledo et al., 2016; Tabassum et al., 2010; Tsolaki et al., 2016).

The friction phenomena at the bone–implant interface are one of the main bases of primary stability (Hansson & Norton, 1999). The idea is similar to stabilization of fractured long bones. In order for adequate healing to occur, there should be no movement between pieces of bone when the ends of a fractured long bone are reduced (Perren, 2002). Primary seating and stabilization of

an implant in the bone are essential for attaining osseointegration. When primary stability is not adequate, this might lead to micromotions (Brunski, 1992). Micro-movements between 100-150  $\mu\text{m}$  can create stresses at the interface, which may affect bone regeneration (Brunski, 1993; Soballe et al., 1993). This jeopardizes the healing process, resulting in fibrous tissue formation and implant failure. In addition, extreme unnecessary stresses applied too early to the bone-implant interface may lead to injury of the bone cells in that area, causing bone necrosis (Atsumi et al., 2007; Buser et al., 2017a; Haiat et al., 2014; Javed et al., 2013; Javed & Romanos, 2010; Mathieu et al., 2014). Therefore, primary mechanical implant stability is an essential ingredient for both short- and long-term clinical success with implant therapy (Buser et al., 2017b; Haiat et al., 2014; Javed et al., 2013; Mathieu et al., 2014; Quirynen et al., 2014).

### *2.2.2 Secondary stability.*

Secondary stability refers to the biologic anchorage mechanism that exists on a cellular level, where bone grows directly onto the surface of an implant. Such stability is a result of bone remodeling and bone apposition connecting the bone and the implant surface (Bosshardt et al., 2017; Buser et al., 2017a; Buser et al., 2017b). Brånemark coined the term “osseointegration,” defining it as “a direct structural and functional connection between ordered, living bone and the surface of a load-carrying implant” (Branemark et al., 1977) (p.3). The osseointegration process takes time, as bone cells regenerate and bone is remodeled, as can be observed histologically and ultrastructurally (Branemark, 1983; Branemark et al., 1985; Linder et al., 1983) i.e. biological stability of implants in bone (Greenstein et al., 2008; Natali et al., 2009). Histologically, osseointegration is similar to functional ankyloses with no intervention of fibrous or connective tissue between bone and implant (Branemark, 1983; Branemark et al., 1985; Linder et al., 1983).



Another definition of osseointegration based on histology is “the formation of a direct interface between an orthopedic or dental implant and bone, without intervening soft tissue” (Miller & Keane, 2003). According to Branemark and Skalak (1997), an osseointegrated implant is able to provide “a stable and apparently immobile support of a prosthesis under functional loads without pain, inflammation or loosening” (p. xi). Moreover, Branemark and Skalak (1997) described osseointegration as the condition of “no progressive relative motion between the implant and surrounding living bone and marrow under functional levels and types of loading for the entire life of the patient” (p. xi). However, in reality, a small amount of micromotion does exist, as bone is not rigid in structure. In addition, the interface of an osseointegrated implant does not consist of complete bone in contact with the implant. Rather, it incorporates marrow-type tissue, as well (Al-Nawas et al., 2006).

Secondary stability occurs after healing of the bone and tissues surrounding the implant, which takes up to 8 months. When an implant has sufficient secondary stability, it is assumed to be able to support the final prosthetic restoration. However, there is insufficient information about how much secondary stability is adequate to withstand biting forces (Trisi et al., 2016). Secondary stability depends on the primary stability supported by new bone growth around the implant, creating a rigid fixation (Brunski, 1992, 1999; Brunski et al., 1979; Fang et al., 2014). In addition, the degree of osseointegration depends on implant surface characteristics (Abrahamsson et al., 2004; Abuhussein et al., 2010; Bosshardt et al., 2017; Doornewaard et al., 2016; Esposito et al., 2014; Hansson & Norton, 1999; Lin et al., 2009; Shibata & Tanimoto, 2015; Tabassum et al., 2010; von Wilmsky et al., 2014; Wennerberg & Albrektsson, 2009) as well as the amount of bone at the interface (Becker et al., 1999; Bell & Bell, 2014; Donati et al., 2008; Drago, 1992;

Elmengaard et al., 2005; He et al., 2015; Trisi et al., 2015). These factors can alter the time needed to reach an adequate secondary stability (osseointegration) capable of safely carrying occlusal loads (Trisi et al., 2016).

Osteoclastic activity is believed to take place before new bone formation occurs. Consequently, stability tends to drop a few weeks after implantation (Haiat et al., 2014). The key to long-term success for a healed implant is an osseous attachment (Albrektsson & Zarb, 1993; Atsumi et al., 2007; Buser et al., 2017b; Haiat et al., 2014; Mathieu et al., 2014).

The main causes of implant failure are impaired host response, disturbance of a weak bone-implant interface after abutment connection, and postsurgical infection (Esposito et al., 1999b).

## **2.3 Factors Affecting Dental Implant Stability and Osseointegration**

As mentioned above, many factors are thought to influence primary dental implant stability, leading to the formation of a strong osseous attachment and long-term implant success. In this section, we will focus on dental implant related factors (such as implant design, geometry, and surface characteristics) and bone factors, including bone density.

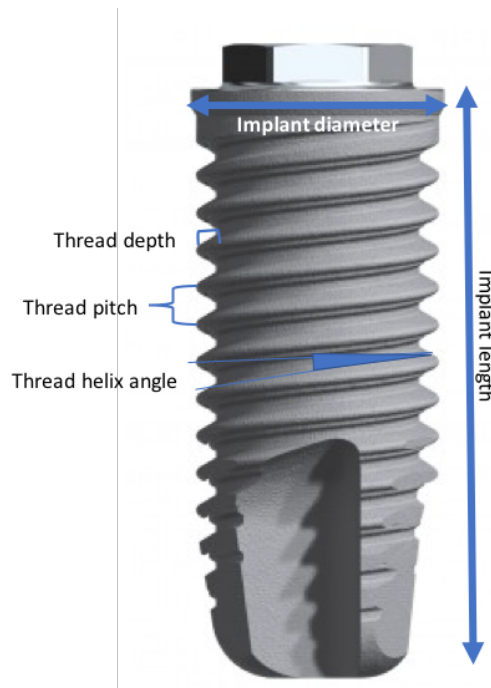
### *2.3.1 Dental implant related factors.*

#### 2.3.1.1 Implant design

Implant design is considered one of the most influential factors affecting implant stability and ability to withstand load. Numerous implant designs have been developed in order to increase the long-term success rates of implant therapy. There are approximately 1300 types of dental implants available in the market, featuring different materials, body shapes, diameters, lengths,

platforms, surface properties, and coatings (Esposito et al., 2014). Implant design has often been influenced by marketing considerations, as opposed to scientific development (Brunski, 1999). Implant manufacturers have often mimicked or made small modifications of other manufacturers' implants. In addition, dental practitioners, in some cases, have used new implants prior to the completion of clinical research (Mathieu et al., 2014). Moreover, because of the disproportionate influence of marketing considerations, it is often difficult to determine whether certain implant surface properties, implant shapes or materials actually enhance clinical outcomes.

Implant design can be classified as either macro-design or micro-design. Macro-design involves thread, body shape, and thread design (e.g., thread geometry, thread pitch, thread depth/height, thickness/width or thread helix angle), as illustrated by Figure 2.1. Micro-design includes implant materials, surface morphology, and surface coating (Abuhussein et al., 2010; Geng et al., 2004a; Geng et al., 2004b). Most current endosseous implant systems are in a tapered shape or available as solid or hollow screws or cylinders. Implant manufacturers provide implants in different shapes, for different conditions. The majority of implants available for clinical use is threaded screws. The original Branemark implant had a machined surface and a parallel design. These implants were not ideal for every case, as they require longer healing time and have a lower survival rate for implants  $\leq 8\text{mm}$  in length (Javed et al., 2013). Tapered implants were later introduced, to achieve better aesthetics and to assist in placing implants adjacent to natural teeth (Shapoff, 2002). Tapered implants are thought to compress cortical bone in areas of inadequate bone, so as to provide better stability (O'Sullivan et al., 2004b).



**Figure 2.1** The implant macrodesign of the Branemark MKIII (NobelBiocare, Switzerland), showing implant length, diameter, thread pitch, thread width, and thread helix angle.

Thread geometry has a positive effect on implant biomechanics. Thread shapes available include: V-shape, square shape, buttress, and reverse buttress shape (Boggan et al., 1999). Threads are used to maximize initial contact, improve primary stability, and enhance implant surface area (Ivanoff et al., 1999). Implant threads should deliver optimal favorable stresses, while reducing extreme adverse stresses to the area of bone-to-implant contact. In addition, implant threads should provide stability and enhance implant surface contact area. Moreover, stress distribution along the implant should be even and minimal, in order to avoid possible complications (Abuhussein et al., 2010; Hansson & Werke, 2003; Lee et al., 2005). In a finite element study by Geng et al. (2004a),

v-thread and large square-thread designs were observed to distribute stresses evenly. Thin thread designs, in contrast, experienced large stress concentrations on the surrounding bone. Another finite element study demonstrated that thread designs do not affect stresses distributed in bone (Eraslan & Inan, 2010). In a study involving animals, square-thread implants experienced significantly higher bone-to-implant contact and torque removal than v-thread and buttress designs (Steigenga et al., 2004).

Thread pitch (see Figure 2.1), the number of threads per unit length, is an important factor in implant design. Increasing the pitch and depth between implant threads may enhance contact between the bone and dental implant (Lee et al., 2005). Implants exhibit some bone loss after osseointegration and over time. Micro-threads (retention grooves) at the cervical part of the implant may help distribute stresses, reducing the amount of bone loss after implant placement (Hansson & Werke, 2003). Not many *in vivo* studies discuss the effect of certain aspects of dental implant design (e.g., helix angle, thread depth, and thread pitch) on success rates (Abuhussein et al., 2010).

#### 2.3.1.2 Implant biomaterial and surface roughness

Dental implants are manufactured using different materials, including titanium with various degrees of purity, titanium alloys (e.g., Ti6Al4V) (Altuna et al., 2016; Gottlow et al., 2012), and more recently ceramics (e.g. zirconia) (Buser et al., 2017b; Depprich et al., 2014; Hashim et al., 2016; Manzano et al., 2014; Sennerby et al., 2005). These materials have been studied extensively in the last few years. Titanium remains the material of choice for dental implants, as it offers excellent mechanical properties, high corrosion resistance, and substantial biocompatibility. Many studies have shown titanium to be biocompatible for long-term use in the

oral cavity (Andreiotelli et al., 2009). However, in some cases, titanium may pose a health hazard due to the release of titanium particles and corrosion byproducts (Bilhan et al., 2013; Evrard et al., 2010; Sicilia et al., 2008; Siddiqi et al., 2011; Tschernitschek et al., 2005). High titanium concentrations have been discovered in areas in the mouth close to titanium implants (Bianco et al., 1996) and in adjacent lymph nodes (Weingart et al., 1994). In recent years, ceramics—especially zirconia (yttria-stabilized tetragonal zirconia polycrystal) implants—have been used for improved aesthetics, in patients sensitive to titanium or requesting metal-free implants. Although zirconia has the potential to be a successful implant material, more clinical studies are required to prove its effectiveness (Buser et al., 2017b; Depprich et al., 2014; Hashim et al., 2016; Manzano et al., 2014; Sennerby et al., 2005).

Surface properties have been a major focus of dental implant research in the last 25 years (Bosshardt et al., 2017). Surface modifications have been created to enhance the bone-to-implant tissue response and to increase clinical performance (Coelho et al., 2009). This is based on the idea that rough surfaces improve the union between implant and bone, by providing more surface area for bone to be in contact with, and also increase mechanical retention. Various methods to improve the implant's mechanical retention include turned, machined, blasted, acid-etched, porous-sintered, oxidized, plasma-sprayed, laser modified, micro-rough, nano-rough, hydroxyapatite-coated, and highly hydrophilic (Esposito et al., 2014; Wennerberg & Albrektsson, 2009). The earliest implant systems had unique surface characteristics: Branemark (Nobelpharma, Gothenburg, Sweden) featured a machined or turned mild rough surface (Albrektsson et al., 1981), while titanium plasma-sprayed implants (manufactured by (Straumann, Basel, Switzerland) were characterized by rough micropores (Schroeder et al., 1981). After 20 years of function, Branemark

implants have demonstrated a 85-95% success rate in complete edentulous cases (Astrand et al., 2008; Attard & Zarb, 2004a, 2004b; Ekelund et al., 2003; Lindquist et al., 1996) and a 91.5 % success rate for single-implant cases (Dierens et al., 2012; Dierens et al., 2013). A 20-year prospective study by Chappuis et al. (2013) revealed an 89.5% success rate for titanium plasma-sprayed Straumann implants in partially edentulous jaws. An *in vitro* study by Mazzo et al. (2012) of implants with acid surface treatment had higher values of pull-out strength relative to machined surface implants, although the two types did not differ in terms of insertion torque values. Various studies have pointed out that surface roughness may not influence the stability at placement; rather, increased primary stability might lead to an increase in stability, as surface roughness influences the bone-implant interface months later (Nergiz et al., 2009; Neugebauer et al., 2009). With an increase in surface modifications, however, there was a corresponding increase in the incidence of peri-implantitis, when compared with implants with turned/machined surfaces (Esposito et al., 1999a; Esposito et al., 1997). In a series of retrospective studies by Roos-Jansaker et al. (2006a, 2006b); Roos-Jansaker et al. (2006c) peri-implantitis was reported to affect 16% of machined or turned implant cases after 9 to 14 years of function.

Surprisingly, a Cochrane systematic review by Esposito et al. (2014), based on retrospective clinical trials (RCT), found no evidence showing any type of dental implant demonstrated excellent long-term success. However, limited evidence demonstrated that implants with rough surfaces such as those that were titanium-plasma-sprayed had a higher susceptibility to peri-implantitis than implants with machined or turned surfaces. They also indicated that there are limited RCTs available in the literature. The RCTs available often had a high risk of bias, with a small sample size and short follow-up periods.

#### 2.3.1.3 Implant geometry

The quantity and quality of alveolar bone greatly influence the length, diameter, and shape of an implant that can be selected for a clinical case (Friberg et al., 2002; Miyamoto et al., 2005; Ottoni et al., 2005; Petrie & Williams, 2005). In addition, the space available for prosthetic placement and the type of occlusal forces involved, all play a major role in the selection of a suitable implant design (Degidi et al., 2008).

Implant diameter is measured from the peak of the widest thread to the same point on the opposite side of the implant (Lee et al., 2005), as shown in Figure 2.1. Researchers have tried to classify dental implants according to their width or diameter, as there are no established criteria for classification (Degidi et al., 2009b; Quek et al., 2006; Romeo et al., 2006). Implants with diameters  $<3.75$  mm are considered narrow-diameter implants, those with diameters  $>3.75$  mm but  $<4.5$  mm are considered conventional diameter implants, and those with diameters  $>5$  mm are considered wide diameter (Degidi et al., 2007b; Javed & Romanos, 2015). Wide-diameter implants, introduced to the market in the early 1990s, were indicated for cases with poor bone quality, inadequate bone height, and cases requiring immediate replacement of failed implants (Langer et al., 1993). These wide-diameter implants, provide greater bone-to-implant contact, counteracting insufficient bone quantity. Because of the presence of scarce bone and reduced blood supply, wide implants may have negative clinical outcomes (Anner et al., 2005; Ivanoff et al., 1999; Mordenfeld et al., 2004). Implant diameter and geometry have been reported to have unique effects on stress distribution in the cortical plate (Baggi et al., 2008; Santiago et al., 2013). As cortical bone is involved in primary stability more than trabecular bone, implant diameter may contribute more than implant length to primary stability (Bosshardt et al., 2017). Small and Tarnow



(2000) demonstrated that the use of wide-diameter implants ( $\geq 5$  mm) adds stress to the buccal bone, leading to bone resorption and gingival recession. On the other hand, a finite element study by Santiago et al. (2013) indicated that bucco-lingual loading of wide-diameter implants enhanced stress distribution to the surrounding bone, similar to the situation where implants are loaded occlusally. However, Haas et al. (1996) reported that implant diameter and length had no significant effect on the cumulative survival rate (CSR) of implants.

A systematic review by Javed and Romanos (2015) concluded that the role of implant diameter on the long-term survival of dental implants placed in posterior maxilla is secondary. They stated that a good surgical protocol, with successful primary stability at the time of implant placement, in combination with pre- and post-surgical maintenance visits, are more important with regard to enhancing the long-term survival of dental implants.

Narrow-diameter implants are used in cases of limited bone space. Success of these implants depends on careful case selection, successful implant seating, and atraumatic surgical technique, as well as adequate biomechanical conditions and bone density (Olate et al., 2010). Moreover, studies have shown that narrow-diameter implants ( $< 3.0$  mm) offer sufficient primary stability in cases with reduced bone volume (Degidi et al., 2009a; Hallman, 2001). However, narrow implants 3.3 mm in diameter have 25% less resistance to fracture, reduced primary stability, and increased risk of overload (Jorneus, 1996; Vigolo & Givani, 2000). A clinical study by Spiekermann et al. (1995) demonstrated a 91% success rate for 3.3 mm cylindrical implants and a 95% survival rate for 4.0 mm cylindrical implants. Likewise, in a 3-year follow-up study by Andersen et al. (2001), the success rate for 3.25 mm self-tapping titanium implants was 93.8% and 100% for 3.75 mm implants.

Implant length is measured from the platform to the apex of implant (see Figure 2.1). Dental implant length is usually classified as short (6-9 mm), medium (10-12 mm), or long (13-18 mm). Increasing dental implant length is believed to result in increased dental implant stability and overall bone-to-implant contact, assuring a higher survival rate (Lee et al., 2005). Increased implant length can also help distribute torque or shear forces when abutments are screwed into place. Nevertheless, increased implant length does not reduce stresses around the implant at the crest of the ridge (Steigenga et al., 2003).

However, a linear relationship between implant length and success rate has not been confirmed. The limited number of studies focused on dental implant length and its survival rate contain significant disagreements (Georgiopoulos et al., 2007; Kotsovilis et al., 2009). Some studies have demonstrated that short implants had a lower survival rate than long implants (Feldman et al., 2004; Herrmann et al., 2005; Naert et al., 2002; Weng et al., 2003; Winkler et al., 2000; Wyatt & Zarb, 1998). On the other hand, other studies have shown that implant length did not influence survival rate (Degidi et al., 2007a; Feldman et al., 2004; Lemmerman & Lemmerman, 2005; Renouard & Nisand, 2006; Testori et al., 2001). The findings from the latter group of studies are influenced by factors such as the implant's primary stability, the clinician's level of experience, and bone density.

Some of the studies that found a higher failure rate with short implants used a surgical protocol independent of bone density (Jemt & Lekholm, 1995; Tada et al., 2003; Wyatt & Zarb, 1998). A retrospective study by Olate et al. (2010) demonstrated that 6-9 mm short implants have a 0.16 times increased chance of early failure. They also reported that in the long run, an implant's length may be more significant than the implant's diameter, as vertical bone loss exists before

implant loading, and can continue at a rate of 0.2 mm per year. Therefore, after years of function, an implant could lose contact with bone incrementally. They also noted that the lower survival rate for short implants may be related to the clinician's level of experience. A study by Eckert et al. (2001) on single-implant restorations did not observe a relationship between implant length and survival rate, especially for implants over 13 mm in length. Douglass and Merin (2002) and Lum (1991) reported that they found no evidence of any relationship between primary stability and implant length. Moreover, increasing implant length may only increase the success rate to a certain extent. In an animal study by Block et al. (1990) no significant difference was observed between the pullout force and different implant diameters. However, 4-mm-length implants showed significantly lower forces of extraction than the longer 8- or 15-mm implants. Feldman et al. (2004) reported that short ( $\leq 10$  mm) machined-surface implants had a 91.5% cumulative survival rate (CSR) and standard machined-surface implants had a significantly higher CSR (97.7%). The authors also reported that short dual acid-etched implants had better outcomes than machined-surface implants. In a retrospective study (Krennmair & Waldenberger, 2004), 114 patients were followed up with for 12 to 114 months. Narrow, standard, and wide diameter implants  $\geq 10$  mm in length were used in this study. Only 2 maxillary implants failed, and no difference was seen between the different implant diameters. Similar success rates have been reported in other studies (Mangano et al., 2014; Manso & Wassal, 2010; Oliveira et al., 2012).

The controversy observed in the success/failure rates of dental implants in clinical research may be understood at least in part due to the fact that studies sometimes do not provide clear descriptions of the type of implants used (short or wide, for example, or surface properties). In addition, the clinician's level of experience with different implant designs, especially new designs,

affects the outcome of these studies (Renouard & Nisand, 2006). An early numerical study showed that occlusal forces are basically concentrated at the crestal bone, rather than evenly distributed over the entire surface area of the bone-to-implant interface (Lum, 1991). This may be a reason why implant length is not linearly related to biomechanical stability. In addition, short implants are mainly used in areas of poor bone quality, such as the maxillary posterior areas, which might increase failure rate. Moreover, short dental implants have a small surface area and a reduced crown-to-implant ratio, which could increase the applied stresses on crestal bone or cause bone resorption, resulting in reduced osseointegration (Kong et al., 2008).

In the presence of good bone density, Horiuchi et al. (2000) and Griffin and Cheung (2004) recommended that implants be at least 10 mm in length, in order to ensure a high success rate. However, Chiapasco et al. (2001) suggested that implants longer than 14 mm in length and wider than 4 mm in diameter provide better success rates. Kong et al. (2008) suggested the use of long dental implants in areas of low bone quality, in order to reduce the stresses applied on the bone. Li et al. (2011) conducted a numerical study for the purpose of assessing differences in dental implant diameter, ranging from 3-5 mm, and 6-16 mm in length, to verify ideal ranges in a posterior mandibular model of poor bone quality. They concluded that stresses applied on poor quality bone are influenced by both implant diameter and length. Implant diameter had a more significant role in reducing cortical bone stress and increasing implant stability, while implant length was significant in reducing trabecular bone stress under both axial and bucco-lingual loads. Consequently, the ideal combination for a screwed implant was determined to be a diameter exceeding 4.0 mm and implant length exceeding 12 mm in the posterior mandible with poor bone quality (Li et al., 2011).

### 2.3.2 *Bone related factors.*

Bone quality is not defined clearly in the dental literature (Marquezan et al., 2012). Many researchers use bone quality as synonymous with bone density, which is incorrect (Lindh et al., 2004; Molly, 2006). More factors are involved with regard to bone quality. Bone quality involves physiological and structural characteristics, bone metabolism, cell turnover, maturation, intracellular matrix, vascularity, and the amount of bone tissue mineralization (Bergkvist et al., 2010; Lindh et al., 2004; Molly, 2006). These factors may enhance implant therapy, but the role of each one of these factors remains poorly understood (Bergkvist et al., 2010; Molly, 2006).

Bone quantity is a matter of the amount of cortical and trabecular bone (with topographic relationship of significant importance) in which the recipient site is drilled. The density of jaw bone available for the implant influences the healing and overall outcome of dental implant treatment (Branemark et al., 1977; Drago, 1992). Therefore, it is important to evaluate bone density prior to implant placement. Bone density is classified and evaluated in many ways.

The classification proposed by Lekholm et al. (1985) is the most widely accepted classification for bone density (Marquezan et al., 2012). This is based on the radiographic evaluation of the ratio of cortical to trabecular bone. The density of bone is quantified as types I-IV, whereby type I involves nearly the entire jaw comprised of homogenous cortical bone. Type II describes a thick layer of cortical bone surrounding a core of dense trabecular bone. Type III refers to a thin layer of cortical bone surrounding a core of dense trabecular bone. Type IV, in contrast, describes a thin layer of cortical bone surrounding a core of trabecular bone that is low-density in nature (Jeong et al., 2013; Lekholm et al., 1985). This system of classification, however, is actually highly subjective, as the determination of characteristics is based on visual assessment.

The Misch (1989, 1990, 2008) classification system, on the other hand, is based on the tactile drilling sensation of bone during implant placement. This system is comprised of four bone density groups (D1 to D4), varying in both macroscopic cortical and trabecular bone types, in all regions of the jaw. The classifications break down as follows: D1—dense hard cortical bone, D2—porous cortical bone, D3—course trabecular bone, and D4—fine trabecular bone. This system of classification relies heavily on the clinician's level of experience and is also highly subjective, as pre-surgical evaluation of bone lacks a strictly objective set of criteria (Norton & Gamble, 2001). Nonetheless, both the Misch classification system and the one conceptualized by Lekholm et al. (1985) allow for at least a rough assessment of the quantity of bone available for implant surgery.

The precise correlation between bone density and implant stability is still not certain, however (Hsu et al., 2013a; Jeong et al., 2013; Misch, 1990; Norton & Gamble, 2001; Seong et al., 2009b; Yang et al., 2002). Low-density bone, often referred to as soft bone, exhibits poor mineralization and limited resistance (Jacobs, 2003; Lazzara et al., 1996). Achieving excellent primary stability in soft bones is challenging. Poor bone density is one of the major factors for implant failure, as this condition may be associated with excessive bone resorption in the healing process, in marked contrast with a condition characterized by higher density bone (Albrektsson et al., 2017; Haiat et al., 2014; Herrmann et al., 2005; Jaffin & Berman, 1991; Johns et al., 1992; Ulm et al., 1999), which naturally enhances implant success rates.

Studies have shown that bone density in the maxilla is lower than the density in the mandible (Devlin et al., 1998; Drage et al., 2007; Gulsahi et al., 2010; Gulsahi et al., 2007). Clinical trials have demonstrated that dental implants placed in the anterior mandible tend to have higher survival rates, compared to those placed in the maxilla, especially in the posterior region (Farre-

Pages et al., 2011; Grassi et al., 2007; Javed & Romanos, 2015; Jemt & Stenport, 2011; Jemt et al., 2011; Miyamoto et al., 2005). Another reason for the difference in success rates is that the posterior maxilla is characterized by lower bone density and a thinner cortical bone—with thicker trabecular bone, compared to the mandible (Jacobs, 2003; Shibli et al., 2013). Turkeyilmaz et al. (2007a) stated that bone density around implants placed in mandibular bone tends to be greater than the bone density surrounding implants placed in maxillary bone. In a clinical trial, Turkeyilmaz and McGlumphy (2008) indicated a strong correlation between bone density and dental implant stability. Miyamoto et al. (2005) showed that dental implant stability is significantly related to cortical bone thickness. In addition, Andres-Garcia et al. (2009) suggested that in order to succeed, implants should be at least 10 mm in length in the mandible and 13 mm in the maxilla. Unfortunately, many studies do not report the type of bone (e.g., new, old or total) in contact with the implant. This can cause confusion, especially in the early phases of the healing process (Bosshardt et al., 2011).

#### 2.3.2.1 Measuring bone mineral density

Bone mineral density (BMD) refers to the amount of bone tissue in a certain volume of bone (Gulsahi, 2011). Assessing BMD of the mandible and maxilla can be a useful tool in implant therapy (Gulsahi et al., 2010). A systematic review by Markezan et al. (2012) concluded that there is a positive relationship between BMD and primary implant stability. The authors also mentioned that the methodological quality and bias of the studies available on this topic should be improved, by providing more evidence. Several methods are used to measure bone density, including densitometric measurements of panoramic and periapical radiographs, and advanced tools such as Dual Energy X-Ray Absorptiometry (DEXA), computed tomography (CT), and

cone-beam computed tomography (CBCT). Micro-computed tomography (micro-CT) is used extensively in research to measure bone density (Bouxsein et al., 2010; Lee & Jasiuk, 2014; Mulder et al., 2004; Tamminen et al., 2011). This is a precise and effective three-dimensional (3D) imaging device that gives detailed information about the density, micro-architectural, and mineralization properties of trabecular and cortical bone in a nondestructive manner, which has the ability to show changes in trabecular construction during aging and postnatal development. It is also capable of displaying the physical properties of trabecular bone, such as the degree and distribution of mineralization (Bouxsein et al., 2010; Mulder et al., 2004; Mulder et al., 2005), as well as the ability to assess bone quality and quantity in experimental studies.

Determining implant stability is challenging not only because of the complex nature of bone tissues but also because the process of bone regeneration takes time (weeks to months). Also, the implant and bone are subjected to mechanical stresses within seconds of implant placement, and the properties of the bone-to-implant interface affected by the interaction between bone tissues and the rough implant surface are difficult to assess (Haiat et al., 2014; Mathieu et al., 2014).

The profusion of different implant systems and designs available can make selection of the most appropriate implant design confusing. In cases with good bone quality and good predictive primary stability, implant design features may not be that important for success. On the other hand, under conditions of reduced bone quality or excessive occlusal loads, enhancing implant surface area contacting the bone by using an implant with a smaller pitch, extra threads, deeper threads, reduced thread helix angle, a long implant, and/or a wide diameter implant may mean the difference between long-term success and early failure. It bears mention that success depends on



more than any single factor but is rather a matter of a number of factors, which collectively may affect the treatment outcome (Abuhussein et al., 2010).

## **2.4 Dental Implant Stability Assessment**

Dental implant stability refers to a measure of the immobility of an implant. Such immobility is a requirement for functional dental implants, and an indirect indicator of the degree of osseointegration. It is critical to evaluate implant stability both before loading an implant and subsequently, on an ongoing basis, to assess the conditions for its success (Haiat et al., 2014). The various techniques and tools used to estimate implant mobility or stability include traditional clinical techniques (primarily non-invasive), experimental tests (invasive), and non-invasive measurement devices.

### *2.4.1 Traditional clinical techniques.*

Dental practitioners have traditionally relied upon the percussion and instrument tests to evaluate dental implant stability. Radiographs are routinely utilized for evaluating conditions relating to the bone-to-implant interface. A percussion test relies on vibrational-acoustic resonance theory, with a metal instrument tapped against an implant to estimate its degree of integration. A clear crystal or ringing sound is an indication of successful integration, while a dull sound may suggest a problem. This technique is obviously very subjective, as it depends on the clinician's level of expertise. Therefore, it cannot be considered a standard tool for determining implant stability (Atsumi et al., 2007; Meredith, 1998a). For tooth mobility assessment, many clinicians use the back handle of two separate instruments simultaneously, tapping on the opposite sides of

an implant abutment to detect implant mobility, to gauge the firmness of the implant. This instrument test, like the percussion test just described, is also highly subjective.

Radiographs, an essential tool for dental implant therapy, are utilized to determine the quantity and orientation of bone, to select implant sites, and to detect the existence of bone disease. An ideal imaging technique would be able to visualize all dimensions of the implant site (mesiodistal, bucco-lingual, and superior-inferior), and to evaluate trabecular bone density, cortical thickness, and to measure accurately. An ideal imaging technique would also be affordable and involve minimal risk of radiation (Benson & Shetty, 2009). No such ideal imaging method exists, unfortunately, as advantages and disadvantages are involved with each tool (Resnik et al., 2008). Traditional radiographs such as peri-apical, bite-wing, and panoramic images have been used for decades to assess implant and bone, especially marginal bone loss (Golubovic et al., 2012; Wang et al., 2013). What is usually used for screening is a widely available panoramic image device, which gives an overview of the maxillofacial area (Benson & Shetty, 2009; Chan et al., 2010b). The main disadvantages of panoramic x-rays are reduced image sharpness and resolution, as well as some distortion. In addition, these images are two-dimensional (2D), showing only mesio-distal (lateral to the implant) bone loss. It is also difficult to predict implant stability and quantify bone density from such panoramic images.

With radiographic technology development, CT and CBCT have gradually become popular tools for dental implant placement, especially in cases of complex reconstruction (Benson & Shetty, 2009; Chan et al., 2010b). CT, which uses Hounsfield units (HU) to define the density of an image, offers several advantages over conventional radiography. CT's high-contrast resolution makes it possible to distinguish tissues that differ less than 1% in physical density. Because CT is

three-dimensional (3D), data from a single scan can be viewed in any arbitrary plane, depending on the diagnostic task. As the density of structures within the CT image is quantitative, it can be used to differentiate tissues and to characterize bone density (Chan et al., 2010b; Gulsahi, 2011; Resnik et al., 2008). The high radiation exposure of CT images, in combination with their relatively high cost and difficulty in accessibility, led to the development of CBCT, which provides accurate, submillimeter-resolution 3D images of the maxillofacial area. CBCT uses pixels and grey level scale, requires less examination time, and may have the capacity to detect bone density but falls short in delivering information regarding implant stability (Soardi et al., 2014). It also has limitations due to its “cone-beam” projection geometry, detector sensitivity, and contrast resolution. The clarity of CBCT images is influenced by artifacts, noise, and weak soft tissue contrast (Scarfe & Farman, 2008). A number of studies have compared the accuracy of CT relative to CBCT in assessing implant site dimensions (Al-Ekrish & Ekram, 2011; Kobayashi et al., 2004; Loubele et al., 2008; Suomalainen et al., 2008). CBCT measurements have been determined to be as accurate as CT measurements (Al-Ekrish & Ekram, 2011; Kobayashi et al., 2004; Loubele et al., 2008; Suomalainen et al., 2008). CBCT voxels are not absolute values, however, as is the case with HU values. Voxels are 3 dimensional pixels, which in CBCT are isotropic voxels (same width, height as depth). Another disadvantage associated with CBCT is the lack of standard for scaling the grey levels representing the reconstructed HU values (Katsumata et al., 2007; Mah et al., 2010; Naitoh et al., 2009, 2010). The algorithms to calculate CT and CBCT are different. Therefore HU cannot be applied to CBCT. These disadvantages notwithstanding, the high radiation exposure of CT images, in combination with their relatively high cost, and difficulty in accessibility, means

that neither system is ideal, making the choice of one over the other a selection of trade-offs, although the overall trend is greater use of CBCT.

The insertion torque (IT) test is derived from the cutting resistance analysis (CRA) that measures the energy ( $\text{J/mm}^3$ ) needed to cut off a unit of bone during implant placement. This energy is significantly correlated with bone density (Friberg et al., 1999a; Friberg et al., 1999b; Friberg et al., 1995a; Friberg et al., 1995b). A torque gauge (N.cm) is usually attached to a hand-piece or torque wrench. The torque gauge reading gives the clinician information about the density of the bone into which the implant is inserted (O'Sullivan et al., 2004a). A high IT value generally indicates that an implant's placement in the bone is firm. Makary et al. (2012) demonstrated that IT agrees well with other measurements such as histologic bone volume. A study by Cannizzaro et al. (2012) reported that when an implant was placed with a torque greater than 35 N.cm, no implant failure was observed. However, 14% of implants failed when they were placed using 25–35 N.cm torque. A major limitation of the IT test is that it cannot determine bone quantity and definitive density at the implant site. Rather, it only gives an objective evaluation of bone density. Moreover, it cannot predict bone loss around an implant or implant failure. IT can be used to estimate primary dental implant stability (Berardini et al., 2016; Haiat et al., 2014) but it cannot be used as an assessment tool for long-term monitoring of implant stability, as it offers only a one-time measurement at the time of implant placement (Haiat et al., 2014).

#### 2.4.2 *Experimental tests.*

Experimental tests—i.e., the reverse torque test, histomorphometric analysis, implant pull-out, and push-in tests—utilize more objective and quantifiable measures, relative to the clinical

techniques mentioned above, but are more destructive and there for used mainly in experimental studies (Atsumi et al., 2007).

The reverse torque (RT) test measures the critical torque threshold at which bone-to-implant contact (BIC) area becomes destroyed (Johansson & Albrektsson, 1987; Johansson & Albrektsson, 1991; Johansson et al., 1991; Roberts et al., 1984). This test offers an indirect assessment about the BIC strength of an implant. In animal studies, the RT value has been associated with histologic BIC measurements (Buser et al., 1998; Roberts et al., 1989; Tjellstrom et al., 1988). In a clinical study, Sullivan et al. (1996) reported that 404 osseointegrated implants had an RT value of 45 to 48 N.cm, with no increase in failure rates. They also concluded that RT at 20 N.cm seems to be a safe, reliable method for verifying osseointegration with pure titanium screw-shaped implants. However, the subjective 20 N.cm threshold used to indicate successful osseointegration needs to be supported by more research. Unfortunately, as a clinical diagnostic tool, the RT test is rather invasive (Meredith, 1998a). The RT value can give some sense as to whether osseointegration is successful, without quantifying the amount of osseointegration. The RT test is therefore now used exclusively in experimental research (Atsumi et al., 2007).

Histomorphometric analysis, generally defined as the measurement of the shape or form of a tissue, is the standard type of measurement used in experimental studies to calculate the amount of bone in contact with an implant surface. Researchers often define osseointegration in terms of percentage of BIC (Hsu et al., 2013b; Tamminen et al., 2011; Watzak et al., 2005). As bone healing occurs around an implant, the BIC may change, indicating that the amount of osseointegration changes with time. For this reason, the BIC may be confusing as a gauge for osseointegration. Nonetheless, investigating BIC offers insight into the local factors influencing stability (Deng et

al., 2008). It can also be used to evaluate the speed of bone formation onto the implant surface (Bosshardt et al., 2017). The substantial sample preparation required is time consuming and expensive, however. Also, histomorphometric analysis is a destructive method that leads to adverse changes or damage to the tissues and implants during preparation. In addition, histomorphometric analysis provides only 2D data, where the histology sections do not represent the entire sample (Elmengaard et al., 2005; Kopp et al., 2012). Therefore, the optimum method for evaluating the bone-implant interface remains controversial (Bouxsein et al., 2010; Schouten et al., 2009; Tamminen et al., 2011).

Micro-CT has been used in dental research as a fast and precise way of analyzing the bone and peri-implant area (Akca et al., 2006; Dias et al., 2015; Fang et al., 2014; Hsu et al., 2014; Hsu et al., 2013b; Kang et al., 2016; Mulder et al., 2004; Neldam & Pinholt, 2014; Park et al., 2005; Schouten et al., 2009; Soardi et al., 2014; Tamminen et al., 2011). Unlike histomorphometric analysis, it does not require the prolonged specimen preparation procedure and is not impaired by the assessment of a limited number of sections. Moreover, micro-CT is a non-destructive method for providing BIC information, with less sample preparation than histology slides. However, the presence of implant artifacts and low-image quality compared to histomorphometric analysis represent major disadvantages for micro-CT (Bouxsein et al., 2010; Dias et al., 2015; Schouten et al., 2009).

Evaluation of the shear strength of the bone-to-implant interface through the use of the pull-out (Schwarz et al., 2009) and the push-in (Ueno et al., 2010) tests is frequently employed for measuring the biomechanical strength of the bone-implant union. Interfacial shear strength is estimated by applying a load parallel to the implant-bone interface, where the implant is either

pulled out or pushed in the bone sample. Such tests, however, are destructive and technique sensitive. They are mainly used with non-threaded cylindrical implants, however, most clinically available implants are of a threaded design. In addition, these tests measure shear stress without any consideration for either tensile or compressive stresses (Brunski et al., 2000).

It becomes evident, then, that conventional methods used to estimate dental implant stability are for the most part subjective and inaccurate. Tests such as histomorphometric analysis may give better information, but are destructive and can only be used in an experimental context. Consequently, there is great clinical need for a non-invasive device with a high level of sensitivity capable of detecting small changes in dental implant stability. A device proficient at measuring smaller changes in implant stability would be helpful in many ways, including: (1) evaluating the implant at different stages of healing on the way to successful osseointegration, (2) assisting critical diagnosis and treatment planning, (3) facilitating the process of decision as to when an implant should be loaded, and (4) monitoring implant status during recall visits.

#### *2.4.3 Current non-invasive methods used to measure implant stability.*

Few of the many devices proposed for measuring dental implant stability are actually available for clinical use. We will start here by explaining experimental modal analysis (EMA) and finite element analysis (FEA), after which we will briefly describe the devices and techniques based on EMA.

EMA is the study of the dynamic properties of a structure under vibrational excitation. It is a standard vibration testing procedure used in mechanical engineering, to simultaneously measure natural frequency (the frequency at which a system tends to oscillate in the absence of any driving or damping force), viscous damping factor (a dimensionless measure describing how

oscillations in a system decay after a disturbance), and stiffness coefficient (the ratio of the force acting on a linear mechanical system, such as a spring, to its displacement from equilibrium). In addition, EMA can detect structural and dynamic changes such as frequency changes, mode, and attenuation of the objects being tested by vibration analysis. It has been used for some time as the standard in medical and dental research (Cunningham et al., 1990; Nakatsuchi et al., 1996a, 1996b; Olsen et al., 2005; Simmons et al., 2001). It also serves as the basis for many dental implant stability devices (Huang et al., 2003a; Huang et al., 2003b; Huang et al., 2002; Huang et al., 2000a; Meredith, 1998a, 1998b). EMA can be implemented both theoretically and experimentally in medical and dental research (Atsumi et al., 2007).

Theoretically, a finite element model (FEM) can be used. In mathematics, a FEM model is used as a numerical technique for solving boundary value problems for partial differential equations. FEM divides a large problem domain into simpler parts, for more efficient solution, reducing related functional error. A two- or three-dimensional FEM with its real boundary conditions can be created by using known biomechanical variables such as Young's modulus (Pa), Poisson ratio and density  $\text{g/cm}^3$  of the object being tested. It can be used to theoretically examine vibrational characteristics of a complex structure and observe its behavior. It can also be used to calculate the expected stress, strain, and other material properties in different simulated models. FEM has the advantage of allowing independent control of each parameter in the models (Atsumi et al., 2007). When estimating dental implant stability, critical analysis is required to identify the determinants of stability. The effect of each factor individually can be difficult to determine, as dental implant stability depends on a number of considerations, including bone and implant properties. Thus, EMA in combination with FEM and simulation is the most appropriate way to

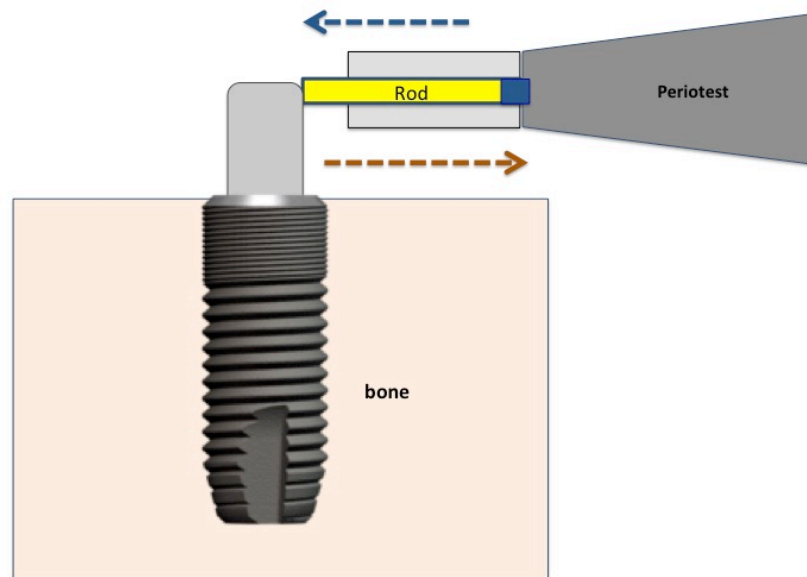


efficiently solve this problem. This, in contrast with animal or clinical models, for which it is considerably more challenging to conduct such studies (Haiat et al., 2014).

The impact hammer method, based on EMA and the percussion test, is used to measure tooth and dental implant stability. As a hammer hits the tooth or implant, the sound from the contact between hammer and object is processed through a fast fourier transformer (FFT). Examples of this technology include: the Dental Mobility Checker (DMC) (J.Morita, Suita Osaka, Japan)—developed by Aoki and Hirakawa (Aoki, 1987; Hirakawa, 1987; Matsuo et al., 1989)—and the Periotest<sup>®</sup> (Medizintechnik Gulden, Modautal, Germany).

The Periotest<sup>®</sup>, developed to quantitatively measure tooth mobility, is one of the earliest devices introduced to the market (Schulte et al., 1983). This device consists of a hand piece with a metal rod that taps on an implant-abutment or tooth to initiate impact. The rod has an accelerometer that measures the acceleration response of the impact. The accelerometer signal is then used to record the time the rod is in contact with the sample being measured as illustrated in Figure 2.2. The Periotest<sup>®</sup> measures the contact time between the rod and the tooth to calculate a Periotest<sup>®</sup> value (PTV). The PTV, which ranges between -08 and +50, has also been used to assess implant stability. Basically, the lower the PTV, the more stable the implant. A fundamental hypothesis of the Periotest<sup>®</sup> is that stronger periodontal stiffness implies a shorter contact time, and vice versa. Considerable controversy surrounds the Periotest<sup>®</sup>, which some previous studies have nonetheless concluded to be a useful device for measuring implant stability (Kaneko, 1991; Lukas & Schulte, 1990; Manz et al., 1992a, 1992b; Olive & Aparicio, 1990; Schulte & Lukas, 1992; Winkler et al., 2001). More recent reviews (Aparicio et al., 2006; Atsumi et al., 2007; Faulkner et al., 2001; Hobkirk et al., 2006; Salvi & Lang, 2004; Swain et al., 2008), however,

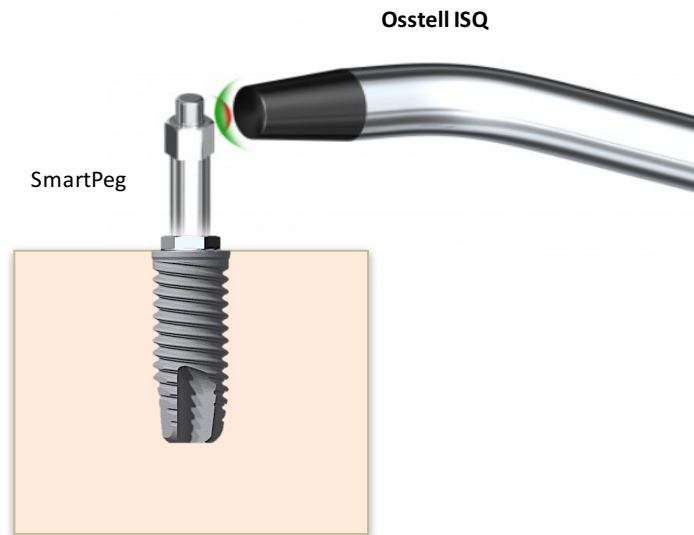
indicate significant limitations, including reduced resolution, low sensitivity, and susceptibility to operator variables (Derhami et al., 1995), which explain the difficulties in acquiring reliable readings from the Periotest<sup>®</sup>. It is also difficult to control PTV-enhancing variables such as implant design, diameter, length, arch location, and bone density (Truhlar et al., 2000). The accuracy of the Periotest<sup>®</sup> for dental implant stability cannot be determined (Salvi & Lang, 2004). It is also unclear exactly how contact time is related to implant stiffness and physical properties of the instrument (e.g., mass of the rod and impact velocity) (Aparicio et al., 2006; Atsumi et al., 2007; Faulkner et al., 2001; Hobkirk et al., 2006; Salvi & Lang, 2004; Swain et al., 2008).



**Figure 2.2** The Periotest<sup>®</sup> consists of a hand piece with a metal rod, used to tap on an implant abutment or tooth, to initiate an impact.

Resonance frequency analysis (RFA), based on EMA, is another non-invasive diagnostic method for measuring implant stability and bone density, at different points in time. The two commercially available devices that currently use RFA are (1) the Implomates (manufactured by Bio Tech One, in Taipei, Taiwan), developed by Huang et al. (Hsu et al., 2013a; Huang et al., 2003a; Huang et al., 2003b; Huang et al., 2002; Huang et al., 2000a; Lee et al., 2000) and (2) the Osstell ISQ<sup>®</sup> (Osstell AB, Göteborg, Sweden) (Cawley et al., 1998; Meredith et al., 1996; Sennerby & Meredith, 1998).

The Osstell ISQ<sup>®</sup> is a non-invasive device used extensively for the assessment of implant stability in experimental and clinical research for over a decade. The device uses RFA to assess implant stability and density of the surrounding bone. A testing peg known as a SmartPeg<sup>®</sup> is hand-tightened to the internal threads of the implant at the fixture level and then stimulated by magnetic pulses produced from the handheld device (Figure 2.3). Using a bending test, the transducer transmits a very low bending force that is transferred as a lateral force to the implant, after which its displacement is measured. A unique SmartPeg<sup>®</sup> is made for each implant design and the abutment of the implant must be removed in order to test the stability of the dental implant (see Figure 2.3).



**Figure 2.3** The SmartPeg<sup>®</sup> is hand-tightened to the internal threads of the implant at the fixture level and then stimulated by magnetic pulses produced by the handheld device (Osstell).

Early Osstell devices were developed as a simple cantilevered bar that could be fixed to an implant fixture or abutment. The bar was stimulated over a range of frequencies, with the first flexural resonance of the resulting system measured in Hz (Sennerby & Meredith, 2008). The rod mounted on the implant had two fundamental resonance frequencies, as it vibrated in two directions, perpendicular to one another. One of the vibrations was in the direction where the implant was most stable, while the other vibration was in the direction where the implant was least stable. Two readings were thus provided, one higher and one lower (Meredith et al., 1996).

In the Osstell ISQ<sup>®</sup>, the natural frequency of the SmartPeg-implant assembly is measured and converted to an Implant Stability Quotient (ISQ) value, ranging from (0 to 100), equivalent to (3,500 to 8,500) Hz. Bornstein et al. (2009) suggested that a value >70 ISQ indicates implant stability, whereas Rodrigo et al. (2010) concluded that a value <60 ISQ indicates that an implant

needs monitoring before loading. A fundamental hypothesis is that the measured ISQ value correlates with the stiffness of the implant-bone system. The many variables believed to influence ISQ values include implant micro- and macro-designs, implant length and diameter (Andres-Garcia et al., 2009; O'Sullivan et al., 2004b; Romanos et al., 2016; Sierra-Rebolledo et al., 2016; Wang et al., 2015), implant positions (Pagliani et al., 2013), bone density (Bardyn et al., 2009; Bilhan et al., 2015; Turkyilmaz & McGlumphy, 2008; Turkyilmaz et al., 2007b; Zix et al., 2005), tightness of the SmartPeg<sup>®</sup>, and its position (Atsumi et al., 2007). Many studies have also demonstrated that RFA is not sensitive enough to detect density of the bone surrounding an implant (Ostman et al., 2006; Scarano et al., 2006). Several studies have indicated that bone contact, especially at the marginal region, plays a critical role in the ISQ reading and is considered a main determinant for RFA measurements (Akkocaoglu et al., 2005; Friberg et al., 1999b). Akkocaoglu et al. (2005) reported that trabecular bone does not enhance ISQ values such as marginal bone.

Researchers have questioned the reliability of ISQ in assessing primary implant stability, as ISQ has been observed as failing to correlate with insertion torque (IT) values (Akkocaoglu et al., 2005; Bernard et al., 2003; Cehreli et al., 2005; Gotfredsen et al., 2000; Kim & Lim, 2011; Krafft et al., 2015; Nkenke et al., 2003; Szmukler-Moncler et al., 2004). Moreover, experimental studies have shown poor correlation between ISQ and histomorphometric measurements such as BIC (Abrahamsson et al., 2009; Ito et al., 2008; Manresa et al., 2014). In addition, a serious clinical drawback of the Osstell ISQ<sup>®</sup> is that the entire complex of screw access whole/prosthesis/abutment must be disassembled all the way down to the fixture, in order to connect the SmartPeg<sup>®</sup>. This is a major problem for a patient with a full-arch multi-implant supported prosthesis. The instrumentation can be rendered useless (and is definitely not non-invasive) if one is evaluating a

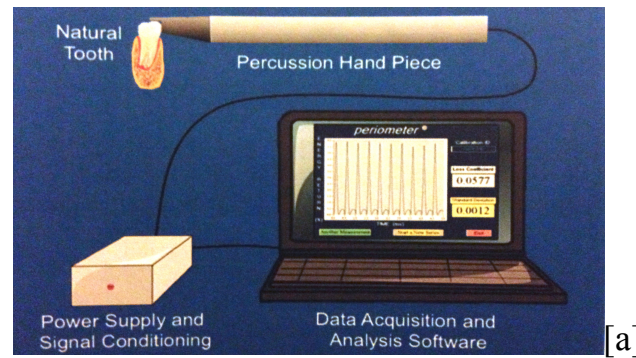
single anterior implant with a cemented crown (Swain et al., 2008). It may also affect the connective tissue and bone around the dental implant, especially in the early stages of healing and osseointegration (Aparicio et al., 2006; Atsumi et al., 2007; Dinh et al., 2013).

The Osstell ISQ<sup>®</sup> is currently being used to evaluate implant primary stability and osseointegration, to determine the possibility of immediate loading, as well as to monitor implant stability over time. A study by Zix et al. (2005) reported that single RFA readings do not give an assessment of the implant's status or predict its outcome. Chan et al. (2010a) demonstrated that the Osstell ISQ<sup>®</sup> was not able to differentiate between implants with and without bone loss. The Osstell ISQ<sup>®</sup> is thought to evaluate an osseointegrated implant and determine the rigidity of the bone-implant interface (Nedir et al., 2004). In cases in which osseointegration is questionable, however, the ISQ value may vary. The Osstell ISQ<sup>®</sup> cannot therefore determine the success, failure, or long-term prognosis of an implant (Feng et al., 2016; Haiat et al., 2014). Clinical studies by Atieh et al. (2014) and Glauser et al. (2004) of immediately loaded dental implants reported a failure rate of 11–28.6% for dental implants with high ISQ values. Thus, the reliability of RFA measurements remains low (Hammerle & van Steenberghe, 2006; Kheir et al., 2016; Lachmann et al., 2006).

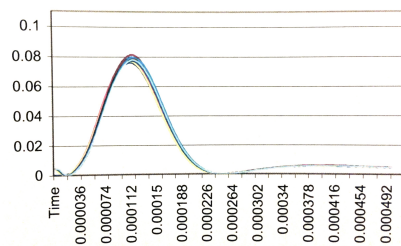
Both the Periotest<sup>®</sup> (Meredith et al., 1998) and the Osstell ISQ<sup>®</sup> (Pattijn et al., 2006) depend on the orientation or fixation of the device, causing reproducibility limitations that may affect consistent quantitative evaluation of implant osseointegration (Vayron et al., 2016). In addition, both devices measure the whole bone-to-implant-abutment/SmartPeg system, not just the bone-to-implant interface (Aparicio et al., 2006), which is a vital parameter for implant stability assessment (Bardyn et al., 2009; Nkenke et al., 2003; Turkyilmaz et al., 2008; Turkyilmaz et al., 2006;

Turkyilmaz et al., 2007b). More research is required to better understand how both the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> estimate implant stability.

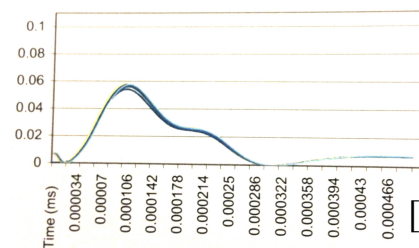
The quantitative percussion diagnostic system Perimeter (Periometrics, Newport Beach, CA). Consisting of a handheld probe with a location pointer and a tip that can be positioned directly against an implant abutment or restoration without disassembly, the Perimeter relies upon the impact response theory known as “damping.” The hand piece is horizontally loaded, with the percussion rod connected to an accelerometer, after which it is activated to impact the sample. The instrument taps on the test abutment with a force of 2-20 N, depending on the material properties of the test sample. Data from the percussion are transferred into raw energy return and displayed on a graph of normalized energy versus time of each impact (see Figure 2.4 [a]). The device then gives the energy loss coefficient value for the sample, a damping capacity parameter indicating the overall energy dissipation in a structure. It normally has values of 0.06-0.09 for natural teeth and 0.03-0.08 for implants. A high loss coefficient value may indicate failing osseointegration, loss of horizontal bone support, poor bone quality or a fractured or loose implant. Then a graph is produced, representing the energy return time profile, A bell-shaped symmetrical curve indicates a healthy tooth or osseointegrated implant (see Figure 2.4 [b]). An abnormal curve, in contrast, would indicate a loose or broken screw, uncemented abutment, loose crown, or bone defects (see Figure 2.4 [c]). The Perimeter also generates a normal fit error reading that characterizes local instabilities in the structure, and calculates the standard deviation, which shows the accuracy of the overall mechanical response of the object being impacted (Dinh et al., 2013; Earthman et al., 2006; Sheets et al., 2013; VanSchoiack et al., 2013). At these early stages of development and marketing, few studies have measured the system’s validity and accuracy (Ahmad & Kelly, 2013).



[a]



[b]



[c]

**Figure 2.4** [a] The quantitative percussion diagnostic system, known as the Perimeter (Periometrics, Newport Beach, CA) [b] A bell-shaped curve indicates a healthy degree of osseointegration. [c] Abnormal curve indicates a problem in a tooth or implant. (Images from Perimeter manual, Newport Beach, CA).

A group of studies by Fung et al. (Feng et al., 2015; Feng et al., 2016; Feng et al., 2014) have recently suggested the use of the damping factor together with RFA in order to determine the properties of the bone-to-implant interface. The damping factor, however, is a difficult parameter to quantify, and can involve a significant degree of error.

Quantitative ultrasound (QUS) (De Almeida et al., 2007; Mathieu et al., 2011a, 2011b, 2015; Mathieu et al., 2014; Vayron et al., 2016) has been recently proposed for evaluating the bone-to-implant interface. More research is needed, however, to improve the reproducibility of the measurements.



Westover et al. (2016) recently developed the Advanced System for Implant Stability Testing (ASIST), which offers a non-invasive means of arriving at a quantitative measurement of implant stability. The ASIST is an integration of a Periotest<sup>®</sup> hand piece with a 4-degree-of-freedom mathematical model. The Periotest<sup>®</sup> hand piece measures the response during the contact, with the linear stiffness at the bone-to-implant interface estimated via a mathematical model. Based on the estimated linear stiffness, an ASIST stability coefficient (ASC) is recorded. The concept was proven through use of a model of bone-anchored hearing aid (BAHA) implants with abutments of various lengths. The study results show that the ASIST can isolate interfacial linear stiffness from other system properties. Although the work by Westover et al. (2016) is pivotal, it is not clear how effectively and accurately the ASIST can extract interfacial stiffness data when applied to dental implants in a porous substrate, such as Sawbones<sup>®</sup>. It is noteworthy that BAHA-implants are significantly shorter than conventional endosseous dental implants. There are questions as to the significant difference in bone-to-implant surface area, non-functional loading versus functional bite forces, and load bearing capacity in conjunction with a chewing force equivalent to hundreds of pounds. Clearly, several major challenges still need to be addressed.

## **2.5 The Different Types of Bone Models Used for Dental Implant Stability Testing *In Vitro***

Dental implants must undergo rigorous testing under both initial *in vitro* and then *in vivo* conditions, to determine the requirements of biocompatibility and mechanical stability prior to clinical use. Animal bone models—e.g., porcine, bovine, canine, and sheep—are usually used for testing dental implants *in vitro* (Pearce et al., 2007; Struillou et al., 2010). Pigs provide a

recommended animal model for experimental studies, since their bone structure is very close to that of humans. Porcine bone is considered representational of human bone tissues, with regard to morphology, bone composition, microstructure, and remodeling characteristics. In addition, porcine bones have similarities with human bones in terms of mineral density and concentration. Porcine bones, however, tend to have a denser trabecular network (Pearce et al., 2007). Although animal models can closely represent the mechanical and physiological clinical situation as it relates to humans, it should be noted that this is only an approximation (Pearce et al., 2007; Struillou et al., 2010). Each type of animal model has unique advantages and disadvantages relative to use as a substitute for humans and should be carefully selected, with the understanding that accuracy of findings as they relate to humans is only approximate (Alzarea, 2014).

Sawbones<sup>®</sup>, artificial bones made of polyurethane plastic, are used to model human bone in many *in vitro* dental implant studies. These artificial bones are designed to simulate bone architecture, as well as the bone's physical properties. Sawbones<sup>®</sup> present appealing advantages compared to human bone: they are easier to store, for example, less expensive, easier to obtain, have no biohazards, and have lower rates of inter-specimen variability in terms of geometry and material properties (MacAvelia et al., 2012). In addition, Sawbones<sup>®</sup> are produced in different foam densities and structures. They can be obtained as a homogeneous test sample or as a hybrid combination of cortical and trabecular bone, to replicate real bones (MacAvelia et al., 2012).

Actual human bones offer unique mechanical and physical properties. However, using actual human bone in biomechanical testing presents some significant challenges. To obtain meaningful and statistically appropriate data, a large number of bone samples is needed. Bones differ greatly in terms of anatomy, geometry, and density, as well as appropriate related

conservation technique. If experiments are to be repeated, therefore (to reproduce results), actual human bone is not regarded to be a uniform test model. Strict laboratory conditions are necessary to demonstrate points of weakness in the technique itself. Considerable advantages notwithstanding, Sawbones<sup>®</sup> do not accurately reflect the biomechanics of human bones. They are used for the purpose of attaining roughly comparable biomechanical results, rather than definite, clinically pertinent, information. In addition to being relatively expensive and difficult to obtain, the use of human bone typically requires approval by ethical committees (Acevedo et al., 2002; Dalton et al., 2003; Hausmann, 2006; Nyska et al., 2002; Richter et al., 2005).

## **2.6 Problems with Studies on Implant Stability *In Vitro***

Many studies have been carried out to assess implant stability, for the purpose of arriving at criteria for clinical use (Bilhan et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016; Sennerby et al., 2015; Tabassum et al., 2010; Wang et al., 2015; Yamaguchi et al., 2015). *In vitro* test models used in these studies has often consisted of dental implants placed in Sawbones<sup>®</sup> (Ahmad & Kelly, 2013; Ahn et al., 2012; Giovanni et al., 2011; Hsu et al., 2013b; Tabassum et al., 2010; Wang et al., 2015). To better control the test models, special attention must be paid to two particular issues: boundary conditions and experimental setup. Boundary conditions are basically the effect of external forces and stresses on a loaded structure and how it is constrained from moving. The test models are subjected to various boundary conditions, so that the test models can be secured and the implants loaded with forces coming from the testing instruments. Ideally, boundary conditions could be chosen, so as not to affect test

results. If not, the boundary conditions should at least be well controlled. Moreover, sensitivity of results to experimental setup should be carefully evaluated. Regardless of how carefully the test models are set up, minute differences between them will nonetheless exist. If the test results were sensitive to the differences in experimental setup, those results might be scattered and inconclusive.

Despite the importance of conditions relating to boundary and experimental setup, the literature includes very few studies addressing these two important aspects. Potential issues relating to experimental setup and boundary conditions for *in vitro* tests have long been ignored, as little discussion has taken place in the dental literature on their effects. Nor is there a standard protocol as to how the conditions relating to boundary and experimental setup should be managed for purposes of *in vitro* dental implant stability testing. One of the few groups of studies to address boundary conditions is that of Huang et al., who conducted a series of *in vitro* (2003a; 2002; 2000a; 2000b) and *in vivo* (2003b) experiments using EMA. The researchers tested the effects of different boundary conditions on natural frequency values, such as clamping force, clamping level (Huang et al., 2000a) thickness of the block, and height of the block, by incrementally decreasing the covering of the implant (Huang et al., 2003a). They also used different bone densities to study the effects on measured natural frequency values (Huang et al., 2002), concluding that the boundary conditions have a significant effect on measured natural frequency values, and that low-density bone reduces the natural frequency values by 90%.

The studies by Huang et al. (2003a; 2003b; 2002; 2000a; 2000b), however, are qualitative. Many of their measured natural frequency values were not evaluated in terms of theoretical predictions. In a few cases, their measured natural frequency values were compared with numerical

results calculated from finite element analyses (FEA), but the agreement was only qualitative (e.g., with regard to trends). The qualitative studies by Huang et al. (2003a; 2003b; 2002; 2000a; 2000b) are certainly adequate for understanding the basic nature and difficulty of measuring implant stability. The lack of quantitative agreement, however, poses a major challenge for reliable assessment of implant stability. The studies by Huang et al. (2003a; 2003b; 2002; 2000) also did not provide benchmarking relative to other tools that assess implant stability (e.g., the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>). It is not clear how effectively other assessment tools assess changes in boundary conditions and experimental setup.

Another problem present in the literature is that measurement of implant stability is generally gauged solely by either the Periotest<sup>®</sup> or the Osstell ISQ<sup>®</sup> (Andres-Garcia et al., 2009; Bergkvist et al., 2010; Bilhan et al., 2015; Bornstein et al., 2009; Chan et al., 2010a; Chong et al., 2009; de Oliveira et al., 2016; Degidi et al., 2015; Ersanli et al., 2005; Lachmann et al., 2006; van Steenberghe et al., 1995; Winkler et al., 2001). As described above, several problems exist with these devices, pointing to the need for use of other standard tests by which to determine viability. Additionally, in many tests that use animal bones for implant stability, the bone density is poorly defined, often based on assumptions rather than direct measurements (Andres-Garcia et al., 2009; Bilhan et al., 2015; Degidi et al., 2017; Kheur et al., 2016).

In summary, implant dentistry needs better clinical diagnostic tools, with greater sensitivity. Current measurement technologies are not well referenced, in the sense that it is not apparent how a particular number that results from the measurement relates to the numbers in other systems. Other issues with current measurement technologies are that they are not highly precise,

have low levels of discrimination, and have little documented correlation to implant health. For example, at what point should a dentist conclude that a certain ISQ value reflects a significant change in the degree of implant stability? The decade-old critique that “the prognostic value of the RFA and Periotest<sup>®</sup> technique in predicting loss of implant stability has yet to be established in prospective clinical studies” (Aparicio et al., 2006, p. 6) remains as valid today as when it was originally made.

Several fundamental questions exist regarding implant stability and measurements: What is the most clinically meaningful way to define implant stability? To what extent is dental implant stability affected by either bone density or the length and width of the implant? How accurate and consistent are the implant stability measurement devices currently in use—and what properties are they actually measuring?

This dissertation offers answers to these questions. The underlying objective is to determine the key biomechanical factors involved in dental implant stability and identify the most reliable and efficient method for testing those key factors. Our experiments will serve to identify the key factors affecting primary dental implant stability. Furthermore, it is the aim here to understand the mechanics by which current dental implant stability devices work. Finally, we endeavor to develop new methods that will serve to measure and detect the stability of dental implants with greater accuracy and efficiency.

The next chapter evaluates the accuracy and reliability of the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter, by calibrating them via experimental modal analysis (EMA) and finite element analysis (FEA). Specifically, measurements by the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter

are compared with those obtained by means of EMA and FEA. We will also analyze the theoretical mechanics used to operate the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter. Rigorous analysis will allow us to understand the assumptions behind and limitations of these devices. The analysis in the next chapter will also provide important clues to how to achieve implant stability by use of these measurements.

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# **CHAPTER 3. CRITICAL ANALYSIS OF IMPLANT STABILITY TESTING DEVICES: BIOMECHANICAL PRINCIPLES AND EXPERIMENTAL DATA**

## **3.1 Introduction**

The preceding chapter described dental implant stability, its types, and the various factors that affect stability. We also described the various methods used to measure dental implant stability (traditional clinical techniques, experimental (invasive), and non-invasive devices). As mentioned in Chapter 2, the two non-invasive devices now commercially available for evaluating implant stability are the Periotest<sup>®</sup> and the Osstell ISQ<sup>®</sup>. The Quantitative Percussion Diagnostics (aka the Periometer) is currently being tested in prototype form.

The objective of this chapter, first, is to evaluate the accuracy and reliability of the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Periometer, by calibrating them via both experimental modal analysis (EMA) and finite element analysis (FEA). EMA and FEA are proven experimental and numerical

methods, respectively, for measuring natural frequency values, viscous damping factors, mode shapes, and stiffness of a mechanical structure. To facilitate the calibration, we placed dental implants of different lengths in synthetic bone models of different densities, and examined how length affects implant stability. Specifically, measurements taken by the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter were compared with those obtained by both EMA and FEA. The second objective of this chapter is to analyze the theoretical mechanics used to operate the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter. The rigorous analysis allows us to understand assumptions and limitations of these devices. The analysis also provides us with important clues as to how to achieve implant stability, using these measurements.

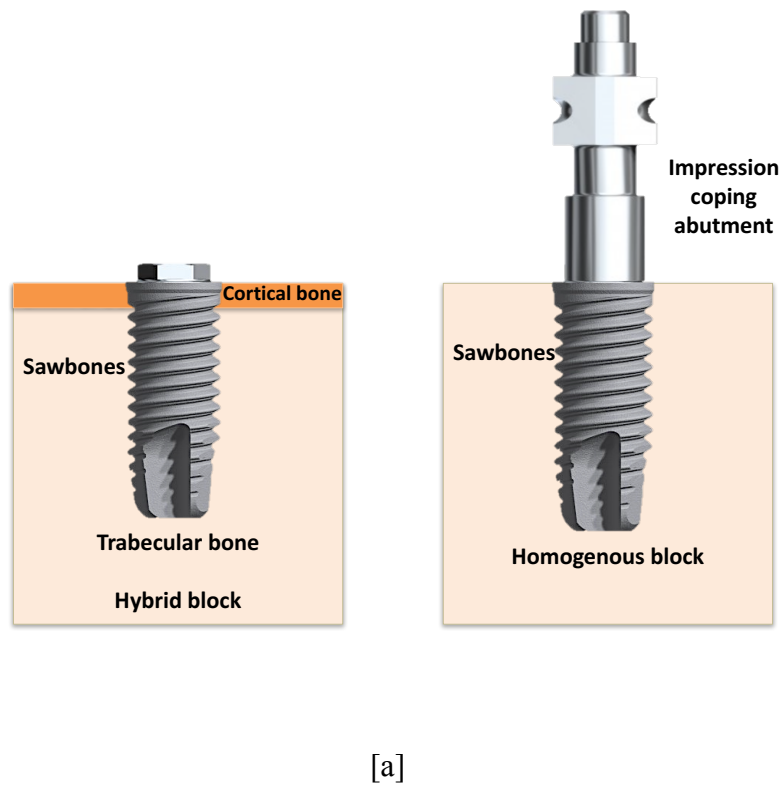
## **3.2 Materials and Methods**

### **3.2.1 Sawbones<sup>®</sup> (Vashon Island, WA).**

Three different Sawbones<sup>®</sup> densities were used: hybrid blocks, high-density blocks, and low-density blocks. Hybrid blocks (34 X 34 X 42 mm in dimension) were used to replicate the average human mandible density (Ahn et al., 2012; Tabassum et al., 2010). They consisted of a 40-mm thick block (20-pound/cubic feet (PCF), 0.32 g/cm<sup>3</sup>) resembling trabecular bone and a 2-mm laminate (40-PCF, 0.64 g/cm<sup>3</sup>) resembling cortical bone. High-density blocks (34 X 34 X 40 mm, 40-PCF, 0.64 g/cm<sup>3</sup>) were used to resemble type I bone according to the Lekholm and Zarb (1985) bone classification system. Low-density blocks (34 X 34 X 40 mm, 15-PCF, 0.24 g/cm<sup>3</sup>) were used to resemble type III-IV bone.

### 3.2.2 Dental implants.

Branemark Mk III implants (Nobel Biocare, Switzerland) 4 mm in diameter and in lengths of 7, 8.5, 10, 11.5, 13, 15, and 18 mm were placed in the center of each block following manufacturer's surgical protocol seating to 45 N·cm as illustrated in Figure 3.1. Branemark dental implants were used in this study primarily because the standard Branemark screw form implant was the first implant design on the market and has been used for more than 45 years. According to the manufacturer, Nobel Biocare, it is the most scientifically documented implant system.



**Figure 3.1** Sawbones<sup>®</sup> models with Branemark Mk III implants: [a] Hybrid laminated block 40/20-PCF (i.e., 2 mm thick cortical layer and lower density 40 mm resembling trabecular bone) and [b] Homogenous block (40-PCF and 15-PCF) with an impression coping (IMP) abutment.

### 3.2.3 *Boundary conditions.*

The lower 13 mm part of each block was firmly clamped to a vise providing as much of a fixed boundary condition as possible. The vise was attached to an isolation table via screws in order to limit external disturbances.

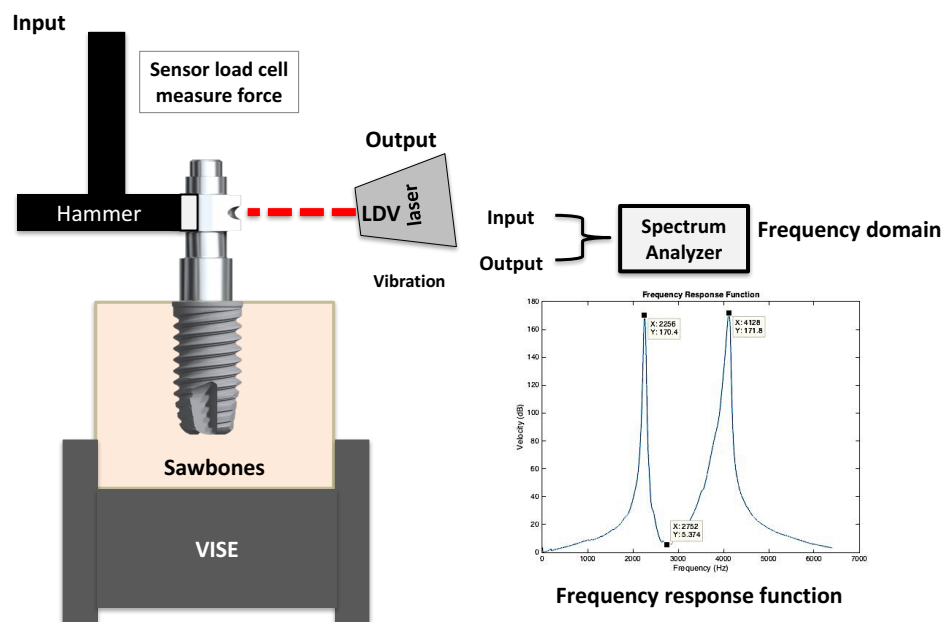
### 3.2.4 *Measurement methods.*

The implant-Sawbones<sup>®</sup> systems were tested using Osstell ISQ<sup>®</sup>, Periotest<sup>®</sup>, Perimeter, and EMA. For the Osstell ISQ<sup>®</sup> tests, SmartPeg<sup>®</sup> (type I) was screwed on the implant at the fixture level with finger pressure. Measurements were taken mesio-distally (MD) and the mean was calculated. For the Periotest<sup>®</sup>, Perimeter, and EMA tests, impression coping (IMP) abutment was screwed at the fixture level (see Figure 3.1 [b]). A 10 N·cm torque was used to tighten the abutments with Branemark System Torque Control (Nobel Biocare, Switzerland). For the Periotest<sup>®</sup> measurements, the samples were tested with the device tip placed perpendicular to the access of the implant and a few millimeters away. For the Perimeter measurements, the samples were tested with the device tip placed on the abutment, perpendicular to the access of the implant. The average of three readings was taken for each set of measurements and the abutment was removed and re-attached between measurements. For the Perimeter, an average of three energy loss coefficient (ELC) readings were taken, where the standard deviation was less than 0.003 for each reading.

### 3.2.5 *Experimental setup for experimental modal analysis (EMA).*

The setup consisted of a hammer (PCB Piezotronics Inc., Depew, NY, USA), a laser Doppler vibrometer (LDV) (Polytec Inc., Dexter, MI, USA), and a spectrum analyzer (Stanford

Research Systems, model SR785, Sunnyvale, CA, USA) as illustrated in Figure 3.2. During the experiment, a hammer tap on the abutment caused the dental implant to vibrate. A load cell (force sensor) on the tip of the hammer measured the force acting on the abutment. Meanwhile, the LDV measured the vibration velocity of the dental implant and its abutment. The measured force and velocity data were fed into the spectrum analyzer, where a frequency response function was calculated in the frequency domain. Various parameters (e.g., natural frequency values, viscous damping factors, and stiffness coefficient) can be extracted from the measured frequency response function. In this chapter, we focused mainly on the natural frequency values and damping ratio.



**Figure 3.2** Setup of experimental modal analysis (EMA) for testing implant-Sawbones<sup>®</sup> assemblies. The setup consists of a hammer (force input) and a laser Doppler vibrometer (velocity output). The measured input and output are processed into a spectrum analyzer to calculate a frequency response function.



### 3.2.5.1 Extraction of natural frequency values.

A natural frequency is one that corresponds to a peak in the measured frequency response function. In general, natural frequency values are very robust quantities to estimate using EMA. They are easy to measure and the measurements are quite repeatable. Since natural frequency values depend on stiffness, their value depends on various factors that affect the stiffness, such as material and geometry of the implants and Sawbones<sup>®</sup>, orientation of the implant with respect to the Sawbones<sup>®</sup>, interfacial properties between the implant and the Sawbones<sup>®</sup>, boundary conditions of the experimental setup (e.g., fixture and how Sawbones<sup>®</sup> blocks are held), and other factors (e.g., residual stresses).

### 3.2.5.2 Extraction of viscous damping ratio.

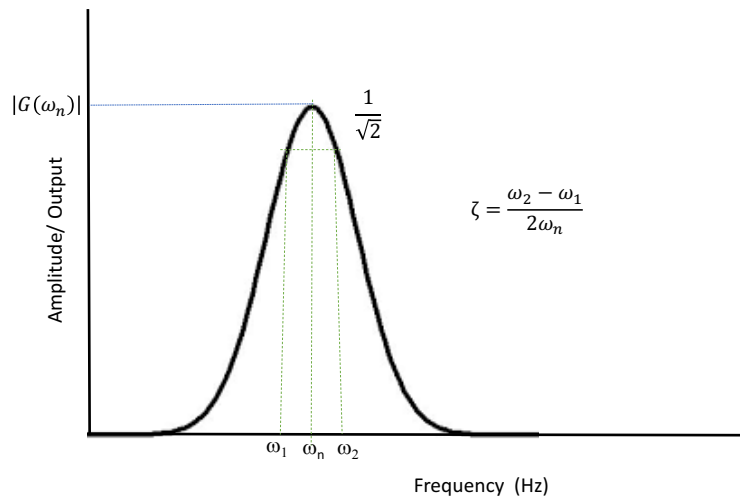
The viscous damping ratio  $\zeta$  can be estimated via the half-power method using the following steps. First, a natural frequency  $\omega_n$  is identified from a peak of the measured frequency response function. The corresponding peak amplitude is  $|G(\omega_n)|$ . Then we can identify two half-power frequencies  $\omega_1$  and  $\omega_2$  satisfying

$$|G(\omega_1)| = |G(\omega_2)| = \frac{1}{\sqrt{2}} |G(\omega_n)|$$

Then the viscous damping ratio  $\zeta$  is estimated via

$$\zeta = \frac{\omega_2 - \omega_1}{2\omega_n}$$

This can be seen in Figure 3.3. It should be noted that damping in general is a very difficult quantity to measure in experiments. No matter which method is used, the measured damping ratio is usually very scattered with a large standard deviation. It is not a robust quantity to measure.



**Figure 3.3** A graph illustrating how viscous damping ratio  $\zeta$  is estimated from a frequency response function using the half-power method. Where  $\omega_n$  is the natural frequency, the corresponding peak amplitude is  $|G(\omega_n)|$  and the two half-power frequencies are  $\omega_1$  and  $\omega_2$ .

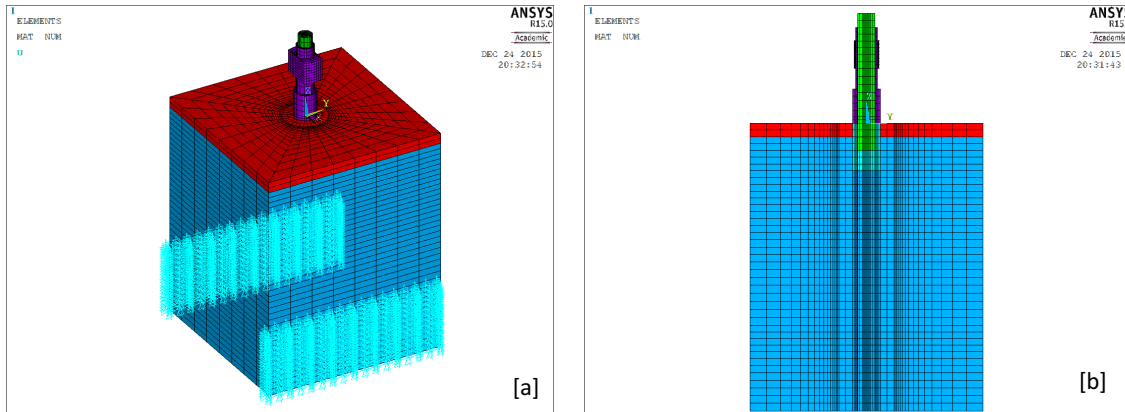
### 3.2.6 Finite element analysis (FEA).

A three-dimensional (3D) finite element model (FEM) was created using ANSYS R-15 (Canonsburg, PA) to simulate the experimental setup and the test results; see Figure 3.4. The model was built using SOLID186 elements, which are higher-order, 3D, 20-node solid elements that assume quadratic displacement fields. The FEM consists of three parts: a Sawbones® block, a cylindrical implant, and an IMP abutment. For the Sawbones® block, material properties (e.g.,

density and Young's modulus) provided by the manufacturer were used to model the tested Sawbones<sup>®</sup> blocks of three different densities (see Table 3.1). The Sawbones<sup>®</sup> block was assumed to be isotropic and has the same size as the test samples. Moreover, the block was fixed at two sides for the lower 13 mm to reflect the boundary conditions imposed in the experiments. To model the implant, one-piece cylinders 7, 8.5, 10, 11.5, 13, 15, and 18 mm in length and 4 mm in diameter were used. Moreover, the cylinders were located at the center of each Sawbones<sup>®</sup> block. The nodes of the cylindrical implants and the Sawbones<sup>®</sup> block were merged at the implant-Sawbones<sup>®</sup> interface to model a no-slip condition (i.e., perfect bonding) between the implant and the Sawbones<sup>®</sup> block. The IMP abutment was modeled in the exact dimensions. It was connected to the implant via a no-slip condition. Titanium was used to model the cylinders and the IMP abutment (see Table 3.1). After the models were created, a modal analysis was conducted to calculate natural frequency values and mode shapes of the simulated test samples.

**Table 3.1** Material properties used for building the finite element models.

| Material        | Density<br>(g/cm <sup>3</sup> ) | Young's modulus<br>(MPa) | Poisson's ratio | Shear strength<br>(MPa) |
|-----------------|---------------------------------|--------------------------|-----------------|-------------------------|
| Sawbones 15-PCF | 0.24                            | 148                      | 0.33            | 33                      |
| Sawbones 20-PCF | 0.32                            | 247                      | 0.33            | 49                      |
| Sawbones 40-PCF | 0.64                            | 879                      | 0.33            | 130                     |
| Titanium        | 4.506                           | 116000                   | 0.32            | N/A                     |



**Figure 3.4** [a] 3D FEM of the fixture level impression coping (IMP) abutment and dental implant embedded in Sawbones<sup>®</sup>, and [b] 2D cross-sectional view of the FEM.

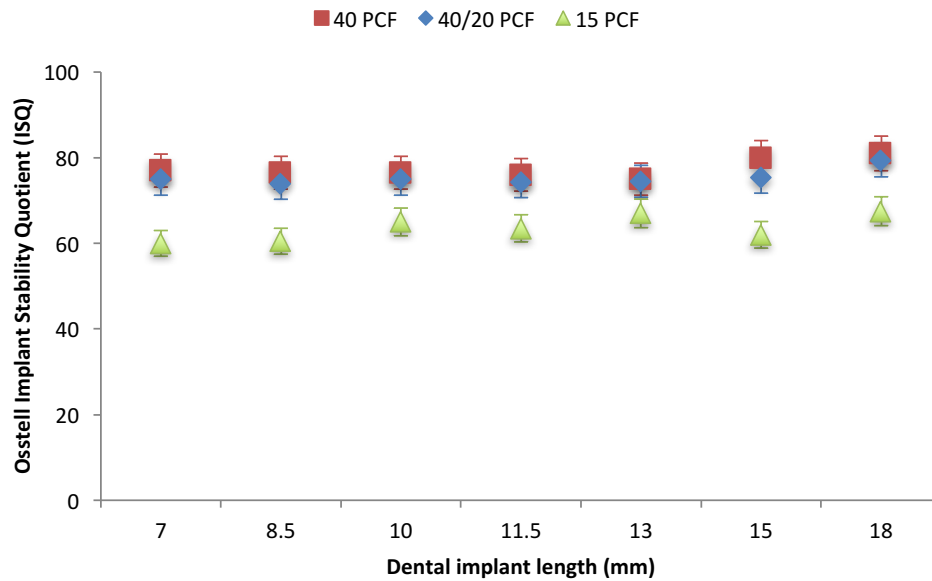
### 3.2.7 Statistics.

The mean ISQ, PTV, ELC, natural frequency values, and damping ratio values were calculated as well as their standard deviations (SD). The correlation coefficient between implant length and these values (ISQ, PTV, ELC, natural frequency values and damping ratio) was estimated.

### 3.3 Results

#### 3.3.1 *Osstell ISQ<sup>®</sup> measurements.*

The mean ISQ MD readings ranged from 75 to 81 for the high-density blocks, 74 to 79.5 for the hybrid blocks, and 60 to 67.5 for the low-density blocks ( $SD \pm 0.5$ ) as seen in Figure 3.5. The ISQ values for the hybrid and high-density blocks fell within the same range, while the ISQ values for the low-density blocks were lower. In other words, the ISQ values for the hybrid blocks were closer to the values of the high-density blocks than those of the low-density blocks. The measured ISQ readings represent implant stability according to the manufacturer. A closer look at the data, however, revealed several subtle observations. First, for the hybrid and high-density blocks, the mean ISQ reading was about the same for implants  $\leq 13$ mm in length. When the implant length was  $\geq 13$  mm, the ISQ reading started to scatter. For the low-density blocks, the ISQ readings were very scattered.

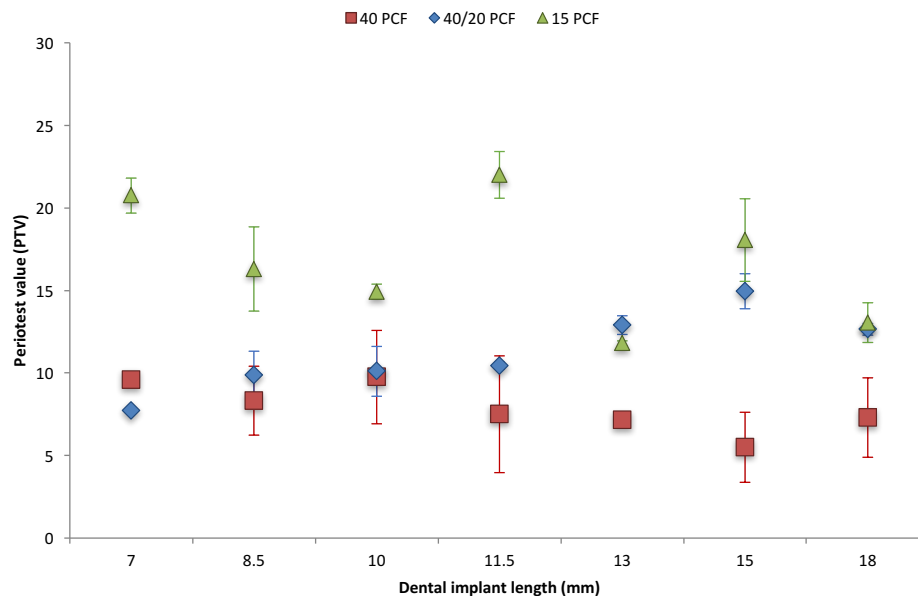


**Figure 3.5** The mean Osstell ISQ<sup>®</sup> values measured for dental implants of ascending length in different Sawbones<sup>®</sup> densities (error bars=SD).

### 3.3.2 *Periotest<sup>®</sup> PTV measurements.*

According to the Periotest<sup>®</sup> guidelines, the lower the PTV reading, the higher the stability. The mean PTV readings ranged between (5.5 and 9.6) for the high-density blocks, (7.7 and 14.95) for the hybrid blocks, and (11.5 and 20.5) for the low-density blocks; the PTV values had high SD as seen in Figure 3.6. In relation to implant length, the PTV for the hybrid blocks had a correlation coefficient of 0.84, implying that the longer the dental implant, the less the stability. For the high-density block group, the correlation coefficient was -0.74 implying that the longer the dental implant, the more stable it is. For the low-density group, the PTV readings had a correlation coefficient -0.47 indicating a slight correlation between dental implant length and stability. In general, the PTV readings were quite scattered and the measurements were inconclusive. The

readings for the different density blocks were haphazard and did not follow a consistent pattern. Nevertheless, PTV readings for high-density blocks were much lower than that of the low-density blocks. According to the Periotest<sup>®</sup> guidelines, the values obtained from these measurements indicated dental implant instability because they were all higher than 0 (see Figure 3.6).

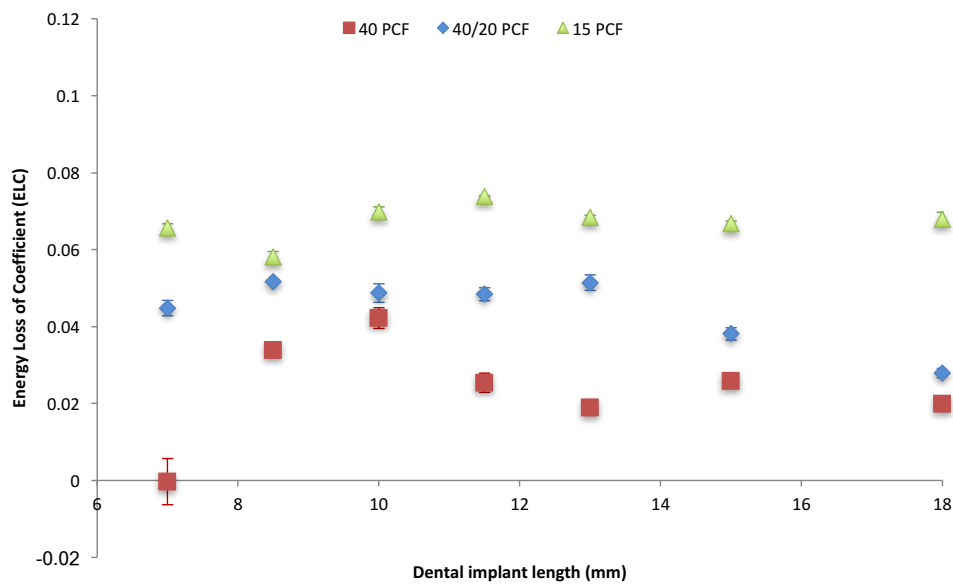


**Figure 3.6** The Periotest<sup>®</sup> values (PTV) measured for dental implants of ascending length in different Sawbones<sup>®</sup> densities (error bars=SD).

### 3.3.3 Perimeter ELC measurements.

According to the Perimeter guidelines, the lower the ELC (energy loss coefficient), the higher the stability of an implant. The mean ELC readings ranged between (0.0581 and 0.0738) for the low-density blocks, (0.0279 and 0.0516) for the hybrid blocks, and (-0.0003 and 0.0422) for the high-density blocks (SD  $\pm$  0.002) as shown in Figure 3.7. In relation to implant length, the implants placed in the hybrid blocks had a correlation coefficient of -0.75, indicating the longer

the implant, the more stable it is. For the low- and high-density blocks, no correlation was found between implant length and ELC. The ELC readings for the different density blocks were inconsistent. However, ELC readings for the low-density blocks were higher than that of the high-density blocks.



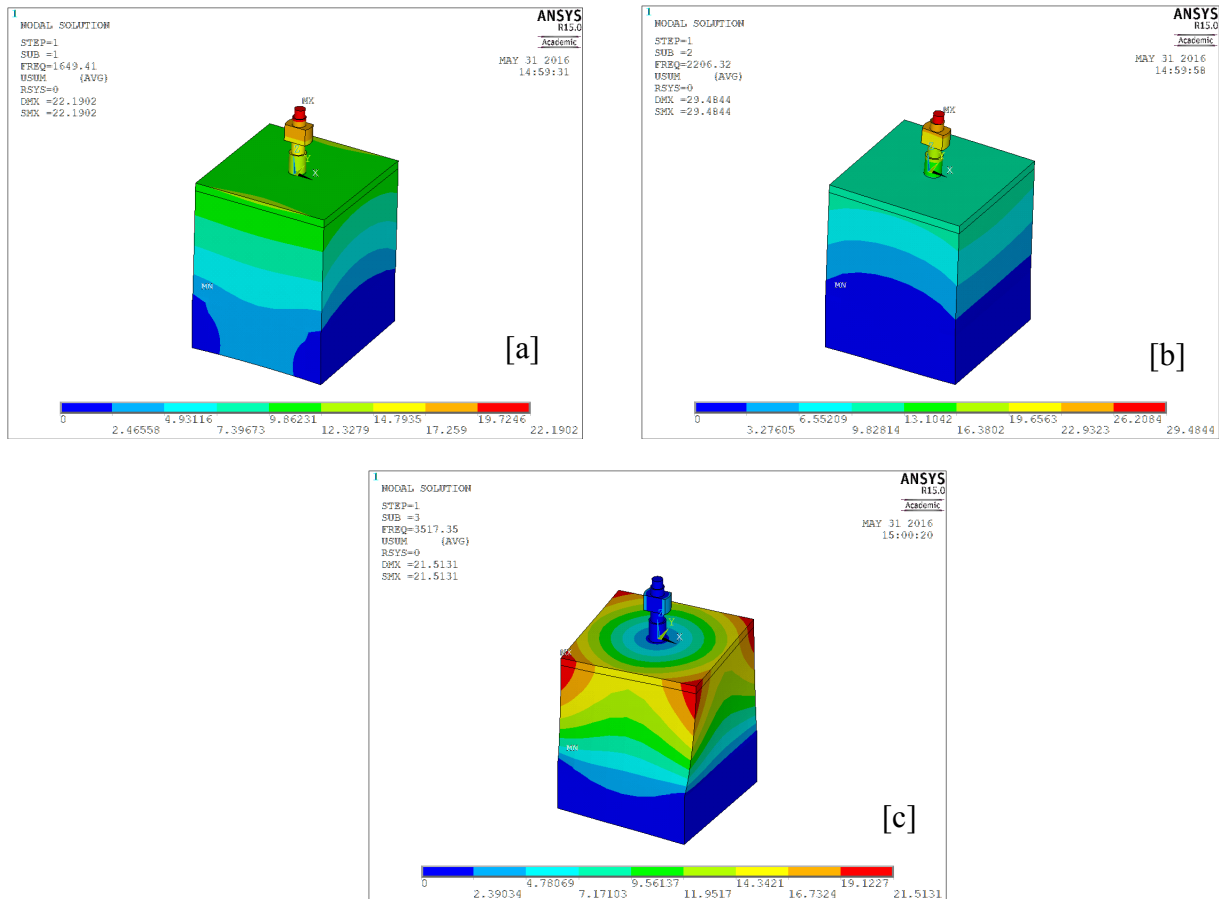
**Figure 3.7** The Perimeter ELC values measured for dental implants of ascending length in different Sawbones<sup>®</sup> densities (error bars=SD).

### 3.3.4 Finite element analysis (FEA).

Three major vibration modes were observed as illustrated in Figure 3.8. The first mode represents a forward and backward movement of the abutment-implant assembly. Since the motion occurs in a direction parallel to the two sides that are partially fixed (as the boundary conditions), the block experiences relatively minor strain leading to less stiffness. Therefore, this mode has the

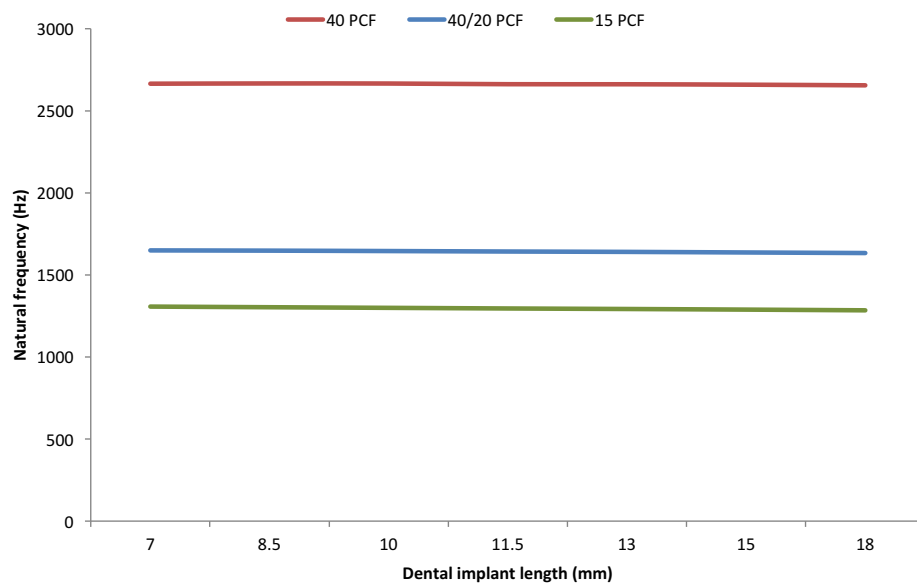


lowest natural frequency. The second mode represents the sideways movement of the abutment-implant assembly. This motion occurs in a direction normal to the two sides that are partially fixed. Therefore, the block experiences relatively larger strain resulting in a higher stiffness and a higher natural frequency. The third mode represents a twisting motion of the Sawbones<sup>®</sup>. Each vibration mode has its own natural frequency value.



**Figure 3.8** The finite element analysis (FEA) results showing three vibration modes. [a] The first mode represents a forward and backward movement of the implant-abutment system. [b] The second mode represents sideways movement of the implant-abutment system. [c] The third mode represents twisting of the Sawbones<sup>®</sup>.

Based on the calculations of the first natural frequency values, there are two major findings. First, the predicted natural frequency values are independent of the implant length as seen in Figure 3.9. Second, the predicted natural frequency values are high for the high-density (40-PCF) blocks and low for the low-density (15-PCF) blocks. Moreover, the difference in frequency values between the 15-PCF and 40/20-PCF (hybrid) blocks is small, while the difference in frequency values between the 40-PCF and 40/20-PCF blocks is more significant.



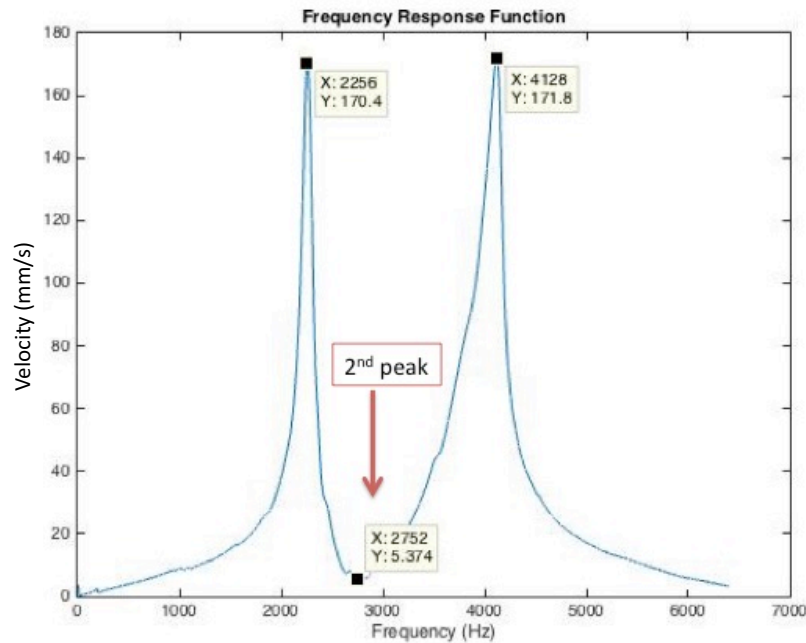
**Figure 3.9** Finite element analysis (FEA) natural frequency predictions based on the first mode.

### 3.3.5 *Experimental modal analysis (EMA).*

The results of EMA had the same trend as the predictions from FEA. Three resonance peaks were seen in the measured frequency response function, confirming the three vibration

modes predicted in FEA. Moreover, the frequency values at which the three resonance peaks appeared were measured natural frequency values. The presence of multiple natural frequency values manifests itself as the complex dynamics of the implant-Sawbones<sup>®</sup> system.

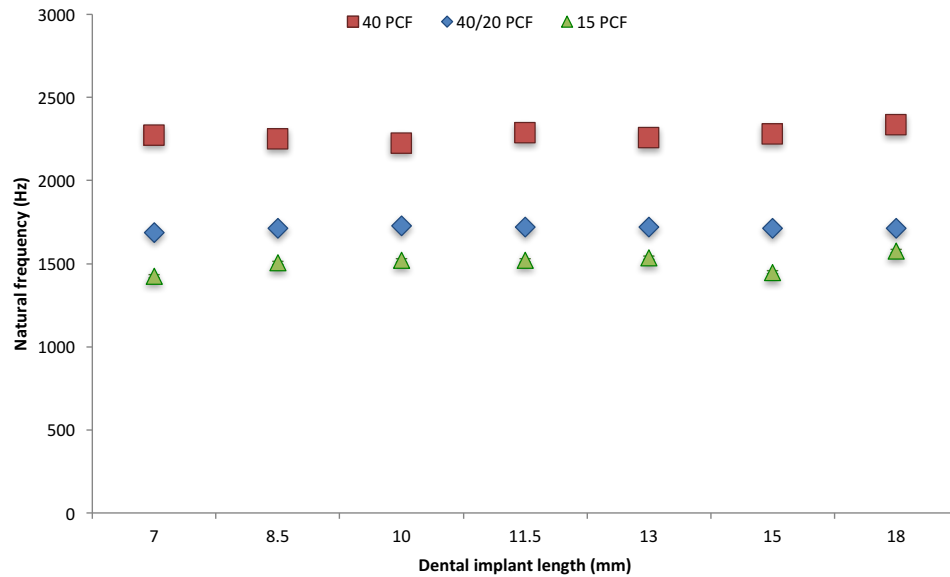
To better excite the first mode, the hammer was adjusted so that it hit the abutment from the front as much as possible. This arrangement minimized excitations from the side and reduced the amplitude of the second peak, and was desirable because we did not want the second mode to interfere with the first mode and contaminate the measured data. As a result, we focused mainly on the first mode, which is the most important mode. When a flawless experiment was conducted, the second peak could not be easily seen because the second mode was not excited, as illustrated in Figure 3.10.



**Figure 3.10** A typical frequency response function of the abutment-dental implant system when the second mode is not excited. Three distinct peaks representing three different modes are expected. The second peak is negligible in amplitude because the abutment is contacted on the broader surface. Illustrated in Figure 3.8, the first mode (i.e., the forward-backward mode) is excited much more than the second mode (i.e., the sideways mode). If we were to activate the system from the side, we would see a more pronounced second peak and a much smaller first peak.

The measured natural frequency values were very consistent across the blocks (maximum  $SD \pm 20$ ). In general, the measured values of the first natural frequency ranged from 2224 to 2336 Hz for the high-density blocks, 1688 to 1720 Hz for the hybrid blocks, and 1424 to 1576 Hz for the low-density blocks (see Figure 3.11). Most importantly, the measured natural frequency values did not vary considerably with respect to implant length as predicted in FEA. The measured frequency difference between the hybrid and the low-density blocks was much smaller than that

between the hybrid blocks and high-density blocks. These results make sense because the low-density blocks are 15-PCF while most the hybrid blocks are 20-PCF; the hybrid-blocks should perform more like the low-density blocks.

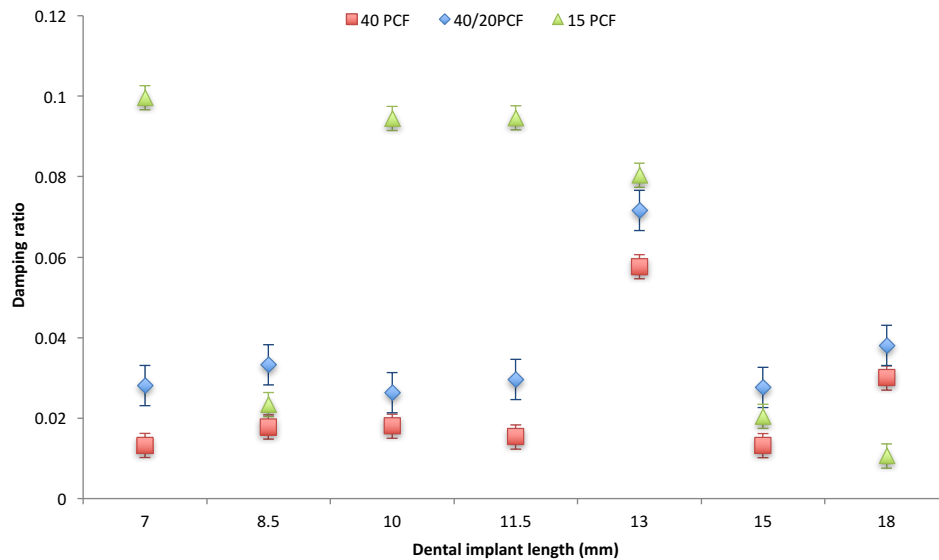


**Figure 3.11** Natural frequency values from experimental modal analysis (EMA). The results followed the predicted finite element analysis (FEA) trend.

These experimental results agree with the predictions from the FEA not only qualitatively but also quantitatively; see Figure 3.8. It should be noted that Sawbones<sup>®</sup> has a  $\pm 10\%$  tolerance in physical properties, which could subsequently affect the measured natural frequency values.

Viscous damping ratios were extracted from the first natural frequency peak as explained above and results are seen in Figure 3.12. The estimated viscous damping ratio values were expected to be high for low-density blocks and low for the high-density blocks, with hybrid blocks

values in between. However, the damping ratio values were inconsistent and had a high standard deviation. There was no significant relationship between the implant length and the damping ratio. The Perimeter ELC values did not follow the trend of the damping ratio values.



**Figure 3.12** Estimated damping ratio values from the first natural frequency values of the frequency response function (error bars =SD).

### 3.4 Discussion

In this study, we aimed to compare results from five different avenues: Osstell ISQ<sup>®</sup>, Periotest<sup>®</sup>, Perimeter, EMA, and FEA. The comparison was carried out on implant-Sawbones<sup>®</sup> models of three different densities, while identifying a correlation between the implant length and primary stability in Sawbones<sup>®</sup>. During the study, we asked the following five questions:

1. What is an accurate benchmark for calibration?
2. How does one develop a good experimental setup?
3. Why are the measured results inconsistent?
4. What is the true meaning of implant stability?
5. What are the advantages and limitations of experimental modal analysis?

Our responses to these inquiries are as follows:

#### 3.4.1 *What is an accurate benchmark for calibration?*

Both EMA and FEA are accurate benchmarks to test implant stability for the following reasons: First, EMA is an experimental analysis whereas FEA is a theoretical analysis. Results from both EMA and FEA were consistent with each other not only qualitatively but also quantitatively. Both EMA and FEA showed that natural frequency values do not depend on dental implant length. Both EMA and FEA had the same trend in which a larger frequency difference was observed between the 40-PCF and 40/20-PCF blocks and a smaller frequency difference was seen between the 15-PCF and 40/20-PCF blocks. Numerical values of natural frequency values obtained from EMA were within 20% of FEA predictions (see Figures 3.9 and 3.11). Therefore, the combined use of EMA and FEA qualifies as a rigorous and reliable benchmark to calibrate commercial implant stability devices.

#### 3.4.2 *How does one develop a good experimental setup?*

Developing an experimental model to test implant stability was the most challenging part of this dissertation. There are no guidelines available for testing dental implant stability *in vitro* and specifically in Sawbones<sup>®</sup>. It was not clear how Sawbones<sup>®</sup> blocks would behave and how to

ideally set up the implant-Sawbones<sup>®</sup> models. Would it be better to place implants in single blocks or all in one block? What would be a proper way to clamp these blocks and stabilize them? What would be the appropriate size or dimensions for each Sawbones<sup>®</sup> block? Unfortunately, previous studies on dental implant stability have not focused on the experimental setup of the test models (Ahn et al., 2012; Bilhan et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016; Sennerby et al., 2015; Tabassum et al., 2010; Wang et al., 2015; Yamaguchi et al., 2015).

In our study, we used FEA to design the test model. Since EMA is very accurate and sensitive, test results faithfully reflect design changes of the test model (e.g., change of block size and how the blocks are clamped). Through the combined use of FEA and EMA, we learned that the following factors are very important to appropriately set up a test model. First, it is not desirable to have multiple implants placed in a large Sawbones<sup>®</sup> block, because each implant vibration is affected by the presence of neighboring implants. Instead, each implant should be placed in its own test block, preferably in the center. As a result, all implants will have the same setup. Second, dimensions of the test blocks must be the same and the orientation of the abutment must also be the same relative to the Sawbones<sup>®</sup> blocks. Finally, the boundary conditions stabilizing the Sawbones<sup>®</sup> blocks are critical. The boundary conditions must be maintained in the same manner for all Sawbones<sup>®</sup> blocks during the tests.

### 3.4.3 *Why are the measured results inconsistent?*

The test results from Osstell ISQ<sup>®</sup> partially followed the benchmark, whereas the test results from Periotest<sup>®</sup> wildly deviated from the benchmark. Moreover, results from the Perimeter



and the damping ratio estimation did not agree. To better understand the results, it was necessary to analyze the mechanical principles behind the operation of the Osstell ISQ<sup>®</sup>, Periotest<sup>®</sup> and the Perimeter.

#### 3.4.3.1 Analysis of the Osstell ISQ<sup>®</sup>.

The basic concept of Osstell ISQ<sup>®</sup> is to extract the natural frequency of an implant-bone system. Since the natural frequency is related to the stiffness of the implant-bone interface, measurements of the natural frequency may reasonably infer implant stability. A major hypothesis is to model the implant-bone system as a single-degree-of-freedom system. In vibration theories, a single-degree-of-freedom system has an undamped natural frequency  $\omega_n$  taking the form of

$$\omega_n = \sqrt{\frac{k}{m}} \quad (1)$$

where  $k$  is a generalized stiffness and  $m$  is a generalized mass of the system. Note that the generalized stiffness  $k$  comes from any part of the system that deforms, and the generalized mass  $m$  comes from any part of the system that moves and presents an inertial effect. If the undamped natural frequency  $\omega_n$  is measured, the generalized stiffness  $k$  can be estimated if the generalized mass  $m$  is known.

When applied to an implant-bone system, equation (1) must be exercised with caution. If the implant and bone are infinitely rigid, the generalized stiffness  $k$  is reduced to the stiffness of the implant-bone interface. Furthermore, if the implant can be considered as a point mass (e.g., small enough in size), the generalized mass  $m$  is reduced to the mass of the implant. If these conditions are not met, exact expressions of the generalized stiffness  $k$  and generalized mass  $m$

could be rather complicated. For example, the generalized stiffness  $k$  may involve the stiffness of both the bone and the implant.

In Osstell ISQ<sup>®</sup>, a small excitation and sensing unit (i.e., the SmartPeg<sup>®</sup>) is connected to the implant. The SmartPeg<sup>®</sup> increases the inertia and modifies the stiffness of the original implant-bone system. Therefore, the natural frequency is now changed to

$$\omega_n = \sqrt{\frac{k+\Delta k}{m+\Delta m}} \quad (2)$$

where  $\Delta m$  and  $\Delta k$  are the inertia and stiffness changes resulting from the presence of the SmartPeg<sup>®</sup>, respectively. An ISQ number is then obtained from  $\omega_n$  as a way to represent the stiffness  $k$ .

This method, in general, is quite robust, because the natural frequency is an easy quantity to measure. There are, however, a couple of issues worth noting. First, the operating principle rests on the assumption that the implant-bone system can be modeled as a single-degree-of-freedom system before and after the SmartPeg<sup>®</sup> is mounted onto the implant. In reality, every implant-bone system is a multi-degrees-of-freedom system presenting multiple modes and natural frequency values, as shown in the FEA and measured frequency response functions in this study. When two natural frequency values are present and detected by the SmartPeg<sup>®</sup> simultaneously, the Osstell ISQ<sup>®</sup> measurements may not be accurate. The second issue is the conversion from the measured natural frequency  $\omega_n$  to the ISQ number. Ideally, the ISQ number is a representation of the stiffness  $k_{int}$  of the implant-bone interface. As shown in (1) and (2), the conversion from the natural frequency  $\omega_n$  will at best give the generalized stiffness  $k$ , which is not necessarily the interfacial stiffness  $k_{int}$ . Even the conversion to the generalized stiffness  $k$  involves many uncertainties, such

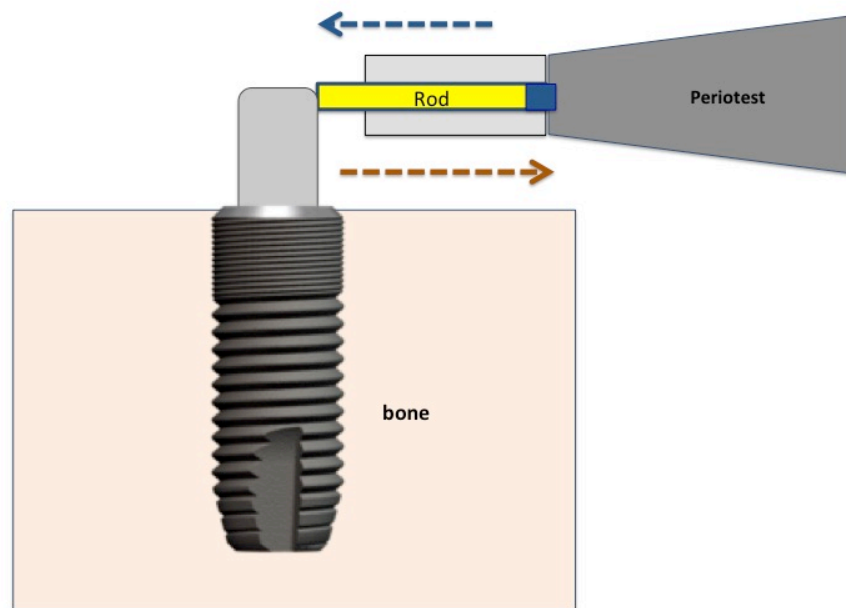
as the exact form of the generalized mass  $m$  as well as the inertia and stiffness effects  $\Delta m$  and  $\Delta k$  from the SmartPeg<sup>®</sup>. Meaningful evaluations of these effects are non-trivial tasks in mechanics.

In this study, analyses of ISQ measurements led to the following observations. First, the measured ISQ values did not show a significant relationship with implant length only when the implant length was  $\leq 13$  mm for both the high-density and hybrid blocks. For this part of the data, the ISQ measurements revealed the same trend observed in the FEA and EMA. For other ISQ measurements, they showed a significant correlation to the increase in dental implant length, as seen in Figure 3.5. Those measurements revealed the limitation of the SmartPeg<sup>®</sup>. For example, the ISQ values for the low-density blocks were very scattered, making the Osstell ISQ<sup>®</sup> measurements less useful. Unfortunately, this is the case in which a dental clinician needs an accurate assessment of implant stability. Second, the Osstell ISQ<sup>®</sup> values for the hybrid blocks were closer to the high-density block values and much higher than the values obtained for the low-density blocks. This is opposite of the trend observed in the FEA and EMA. The measured Osstell ISQ<sup>®</sup> values depend significantly on the cortical layer or on the surface that is closer to or in contact with the SmartPeg<sup>®</sup>. Osstell ISQ<sup>®</sup> could not detect the trend of the three different density blocks. Ahmad and Kelly (2013) observed similar results in their ISQ values. They noted that the Osstell ISQ<sup>®</sup> value was higher than expected when the top half of the implant was embedded in the 40/20-PCF block. Many studies have also demonstrated that resonance frequency analysis (RFA) is not sensitive enough to detect density of the bone surrounding an implant (Ostman et al., 2006; Scarano et al., 2006). Several studies have indicated that bone contact, especially at the marginal region, plays a critical role in the ISQ reading and is considered a main determinant for RFA

measurements (Akkocaoglu et al., 2005; Friberg et al., 1999; Ivanoff et al., 1996). Akkocaoglu et al. (2005) reported that trabecular bone does not enhance ISQ values such as marginal bone.

#### 3.4.3.2 Analysis of the Periotest<sup>®</sup>

In the Periotest<sup>®</sup>, an incoming rod impacts the dental implant. An accelerometer, located on the non-impacted end of the rod, measures the motion of the rod. As the impact begins ( $t = 0$ ), the rod decelerates and rebounds because of the stiffness of the implant-bone system. At time  $t = \tau$ , the rod and the dental implant start to separate. The time duration  $\tau$ , known as the contact time, is used to calculate PTV (see Figure 3.13). Studies have indicated that PTV can present large variations, for example, under various impact positions or angulations (Aparicio et al., 2006; Atsumi et al., 2007; Faulkner et al., 2001; Hobkirk et al., 2006; Kaneko, 1994; Salvi & Lang, 2004; Swain et al., 2008).



**Figure 3.13** The Periotest<sup>®</sup> consists of a hand piece with a metal rod, used to tap on an implant abutment or tooth, to initiate an impact (copied from Chapter 2).

Theoretically, accurate prediction of the contact time is quite complex. Mathematical models can provide insight to the PTV measured. In general, the contact time  $\tau$  is a strong function of stiffness and mass, but a very weak function of damping. Therefore, the Periotest<sup>®</sup> can be used to identify stiffness and is not adequate for measuring damping.

The simplest model is to approximate the impact as an impulsive force. In the Periotest<sup>®</sup>, the incoming rod provides an impulsive force to the implant, and the implant (along with the bone and soft tissue) is modeled as a single-degree-of-freedom spring-mass-damper system as shown in Figure 3.14 [a]. Let us further assume that the generalized mass and stiffness of the implant-bone system are  $m$  and  $k$ , as in the model of the Osstell ISQ<sup>®</sup>. In addition, the mass of the impact rod is  $M$ . When the rod is in contact with the implant, the system will have undamped natural frequency  $\omega_n$  given by

$$\omega_n = \sqrt{\frac{k}{m+M}} \quad (3)$$

Following basic vibration theories, displacement  $x(t)$  of the implant under an impulsive force will then be given as:

$$x(t) = \frac{v_0}{\omega_d} e^{-\zeta \omega_n t} \sin \omega_d t \quad (4)$$

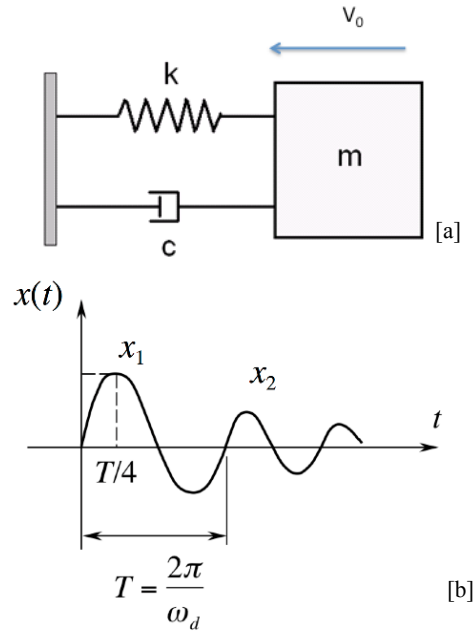
where  $v_0$  is the velocity of the implant after the impact,  $\omega_n$  is the undamped natural frequency of the implant-bone-rod system shown in (3),  $\zeta$  is the viscous damping factor (aka damping ratio), and  $\omega_d$  is the damped natural frequency of the system defined by

$$\omega_d = \sqrt{1 - \zeta^2} \omega_n \quad (5)$$

The typical response given by (4) is plotted in Figure 3.14 [b]. The displacement  $x(t)$  will oscillate and decay as time goes on. The contact time is the time it takes to finish the first half-cycle, i.e.,

$$\tau = \frac{\pi}{\omega_d} = \frac{\pi}{\sqrt{1-\zeta^2}\omega_n} = \frac{\pi}{\sqrt{1-\zeta^2}} \sqrt{\frac{m+M}{k}} \quad (6)$$

As we can see from (6), measurement of the contact time  $\tau$  virtually means measurement of the damped natural frequency  $\omega_d$ . Since the undamped natural frequency  $\omega_n$  is unknown, the contact time  $\tau$  cannot be used to extract damping properties via (5). Moreover, the dependence of contact time  $\tau$  on damping ratio  $\zeta$  is very weak, because  $\zeta$  is in general small and is seldom  $> 0.1$  for dental implants. For example, if  $0 < \zeta < 0.1$ , then  $0.995 < \sqrt{1-\zeta^2} < 1$ . Therefore, damping has almost no effect on the contact time  $\tau$ . The Periotest<sup>®</sup> is used, in principle, to measure undamped natural frequency  $\omega_n$ .



**Figure 3.14** Mathematical model of the Periotest<sup>®</sup>. [a] The model consists of a spring-mass-damper system presenting a single degree of freedom. [b] Displacement  $x(t)$  of the implant under an impulsive force will oscillate and decay as time goes on.

Measuring the damping ratio  $\zeta$ , requires displacement of the dental implant measured for at least two or more oscillations. For example, let the peak amplitude of two consecutive oscillations be  $x_1$  and  $x_2$ . Then a logarithmic decrement  $\delta$  can be defined as:

$$\delta = \ln \frac{x_1}{x_2} \quad (7)$$

and the damping ratio  $\zeta$  can be estimated via

$$\zeta = \frac{\delta}{\sqrt{4\pi^2 + \delta^2}} \quad (8)$$

With the theoretical framework above, it is evident that the Periotest<sup>®</sup> will not provide accurate damping properties for several reasons. First, the Periotest<sup>®</sup> does not measure the motion of the dental implant. Instead, it measures the motion of the impact rod because the accelerometer is on the non-impacted end of the rod. Second, correlating the motion of the dental implant to the motion of the impact rod is only possible during the first half-period when the impact rod and implant are in contact. After that, the impact rod and dental implant separate, and the accelerometer measurements no longer correlate with the motion of the dental implant. Reasonable measurements of the viscous damping factor  $\zeta$ , however, will require at least two oscillations of the dental implant, which is not provided by the Periotest<sup>®</sup>.

A more advanced model is to consider the implant-bone system as an elastic system (e.g., a rigid rod with an elastic support). In that case, the kinetic energy amplitude of the system when the rod is in contact with the implant will take the form of

$$KE = \frac{1}{2} (m + \mu M) v_0^2 \quad (9)$$

where  $\mu$  is a very complex function that depends on the location and angulations of the impact. This kinetic energy corresponds to an effective mass of  $m + \mu M$ , thus modifying the undamped natural frequency to

$$\omega_n = \sqrt{\frac{k}{m+\mu M}} \quad (10)$$

The presence of factor  $\mu$  is the reason that the Periotest<sup>®</sup> PTV results may not be consistent for various locations and angulations. Another problem is that the Periotest<sup>®</sup> may undergo multi-mode interactions. The ability to extract damped natural frequency  $\omega_d$  via (6) is the assumption of a single-degree-of-freedom system. The implant-bone-rod system, however, is not a single-degree-of-freedom system. It has multiple vibration modes and each mode has its own natural frequency. When two or three modes have similar natural frequency values, the contact time  $\tau$  can still be measured via the Periotest<sup>®</sup>. Nevertheless, the interpretation may not be done accurately if the measured contact time  $\tau$  is fitted numerically via (6).

In this study, the PTV readings were not easy to obtain and had a high SD. Every time the rod hit the abutment, it gave a different reading. In general, even being inconsistent, the PTV readings were low for the higher density blocks and higher for the low-density blocks. As for the hybrid blocks, the PTV readings increased with increasing implant length (see Figure 3.6). Most of the PTV measurements did not agree with the results observed in the FEA and EMA.

#### 3.4.3.3 Analysis of the Periometer (quantitative percussion diagnostic system).

Theoretically, the action in the Periometer can be modeled as an elastic rod with an initial velocity  $v_1$  and mass  $m_{rod}$  colliding with a tooth of elastic support (i.e., an elastic object), as illustrated in Figure 3.15 [a]. In modeling such an impact event, one needs to focus on two different time frames. The first-time frame is the moment immediately before the impact, while the second-time frame is any moment after the impact. According to basic principles in mechanics, the following work-energy equation is valid between these two-time frames.



$$(KE)_1 + (PE)_1 = (KE)_2 + (PE)_2 + D_{1-2} \quad (11)$$

Where  $(KE)_1$  is the kinetic energy of the rod and tooth at time frame 1,  $(PE)_1$  is the potential energy of the rod and the tooth at time frame 1,  $(KE)_2$  is the kinetic energy of the rod and tooth at time frame 2,  $(PE)_2$  is the potential energy of the rod and the tooth at time frame 2, and  $D_{1-2}$  is the energy dissipated between time frames 1 and 2. In particular, at time frame 1 before the impact, the kinetic energy of the system is given by

$$(KE)_1 = \frac{1}{2} m_{rod} v_1^2 \quad (12)$$

because the tooth does not have any initial velocity before the impact. Moreover,

$$(PE)_1 = 0 \quad (13)$$

because there is no deformation in the rod or the tooth. After the impact, the rod bounces back and the tooth starts to vibrate. Therefore, kinetic energy at time 2, i.e.,  $(KE)_2$ , will have contributions from the rod and the tooth. The same is true of potential energy at time 2, i.e.,  $(PE)_2$ . The energy dissipated  $D_{1-2}$  is, in general, very complicated. It is conveniently divided as:

$$D_{1-2} = D + D_p \quad (14)$$

where  $D$  is, the energy dissipated by the tooth (e.g., via soft tissue) and  $D_p$  is energy dissipated elsewhere (e.g., energy lost during the contact between the rod and the tooth or energy dissipated in the rod).

After reviewing papers that describe the operation principles of the Perimeter (Dinh et al., 2013; Earthman et al., 2006; Sheets et al., 2013; VanSchoiack et al., 2013; VanSchoiack et al., 2006), we concluded that three major equations are used to develop the theory behind the Perimeter operation:

The first equation is an energy balance given as:

$$D = U - E_\varepsilon - D_p \quad (15)$$

where  $U$  is the input energy,  $E_\varepsilon$  is the potential energy of the rod,  $D_p$  is energy dissipated elsewhere, and  $D$  is the energy dissipated by the tooth.

The second equation is an assumption that the input energy  $U$  is approximated as the kinetic energy of the incoming rod, i.e.,

$$U \approx (KE)_1 = \frac{1}{2} m_{rod} v_1^2 \quad (16)$$

The third equation is a definition of loss factor via

$$\eta = \frac{D}{2\pi U} \quad (17)$$

where  $U$  is the strain energy.

We have observed several major problems in the Perimeter formulations documented in the literature (Dinh et al., 2013; Earthman et al., 2006; Sheets et al., 2013; VanSchoiack et al., 2013; VanSchoiack et al., 2006). The first problem is that major energy terms are missing. By comparing equations (15) and (16) with equations (11) and (14), we can see that kinetic energy (KE) of the rod and tooth after impact is missing in (15), and so is the potential energy (PE) of the tooth. These energy terms cannot be neglected. If they are not included in equation (15), the consequence is that these missing energy terms will be counted as the energy loss  $D$  via equation (15). That means the measured energy loss  $D$  is not the energy loss of the tooth. Instead, it is a combined form of energy loss, kinetic energy, and potential energy. The second problem is that the Perimeter formulation incorrectly uses the input energy  $U$  in (16) and strain energy  $U$  (of the tooth) in (17) interchangeably.

Assuming that the input energy is the same as the strain energy means that the initial kinetic energy is 100% converted to strain energy in the tooth. Theoretically, this would only occur when

the following three conditions are met simultaneously: (a) there is no energy loss, (b) the rod does not deform after an impact, and (c) neither the tooth nor the rod move after an impact. These conditions obviously are not met in real Perimeter tests, which are intended to measure energy loss. The third problem is that  $D_p$  is virtually unknown. To estimate  $D_p$ , the Perimeter uses acrylic and aluminum with known energy loss factors  $\eta_1$  and  $\eta_2$  as a way to calibrate the acrylic and the aluminum blocks, respectively (Dinh et al., 2013; Earthman et al., 2006; R. et al., 2006; Sheets et al., 2013; VanSchoiack et al., 2013; VanSchoiack et al., 2006), by using (15) and (17):

$$2\pi\eta_1 U = U - E_\varepsilon - D_p \quad (18)$$

and

$$2\pi\eta_2 U = U - E_\varepsilon - D_p \quad (19).$$

Then the studies on the Perimeter solve  $D_p$  and  $E_\varepsilon$  from (18) and (19) simultaneously (Dinh et al., 2013; Earthman et al., 2006; Sheets et al., 2013; VanSchoiack et al., 2013; VanSchoiack et al., 2006). The procedure used to estimate  $D_p$  is dubious for many reasons. First, equation (15) is incorrect as already described above. Second, the use of (17) assumes that no energy is lost during an impact. Therefore, the estimation of  $D_p$  is not sound from a theoretical mechanics point of view. Lastly, in order for this calibration to work, it is assumed that  $D_p$  is the same for both aluminum and acrylic; however, there is no evidence to support this assumption. The last problem is curve fitting, since  $E_r$  (energy return) is obtained from experimental measurements, it has a lot of noise. If a noisy signal is used as an input to calculate damping, erroneous results will appear. Therefore, a curve fitting is performed to clean up the noisy signal. Basically, the measured  $E_r$  is fitted into the following expression

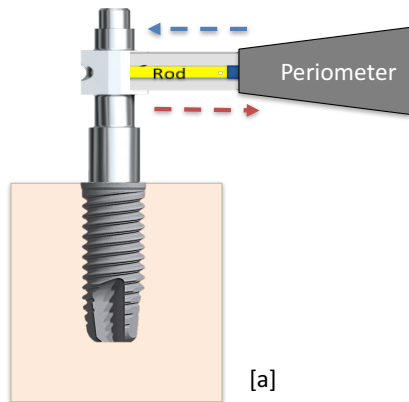
$$E_r = \beta \sin^2(\gamma t) \exp \left[ \frac{(t-\phi)^2}{\psi} \right] \quad (20)$$

Since energy is always positive, the sine function in (20) is squared. In addition,  $\beta$  indicates the amplitude of the energy measured,  $\gamma$  is the frequency of oscillation,  $\phi$  is a phase shift, and  $\psi$  is a time constant characterizing how fast the energy fades with time. By using the curve fitting, one can convert the noisy, measured  $E_r$  into four parameters, i.e.  $\beta$ ,  $\gamma$ ,  $\phi$ , and  $\psi$  (see Figures 3.15 [b] and [c]).

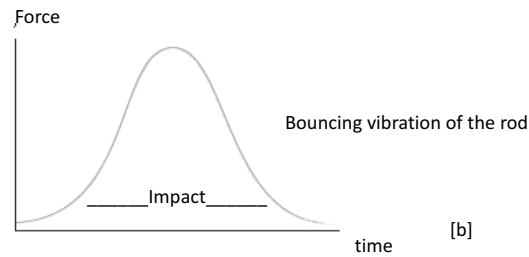
Based on the results from the literature review, there is weak scientific evidence that the Periometer measurements represent damping of the tested implant or tooth.

In the current study, the Periometer ELC readings were very difficult to obtain. Each time the Periometer tapped on the tooth it gave a different number. Overall, even with such inconsistency, the high-density blocks had the lowest ELC values, whereas the low-density blocks had higher values. The hybrid block values decreased with increasing implant length (see Figure 3.7).

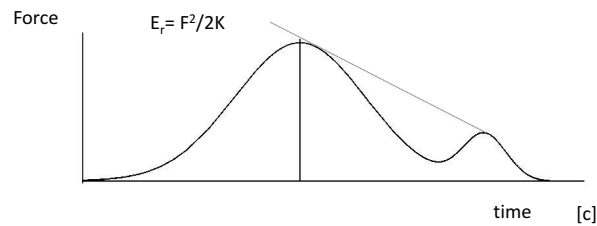
The damping ratio was estimated from the frequency response function of EMA. The results showed that the damping ratio varied with varying bone densities. The damping ratio was higher for low-density blocks and low for high-density blocks, yet we observed a few low-density blocks with lower damping ratios (see Figure 3.12). The damping ratio values had high SD values. Thus, damping ratio is hard to estimate and is not a reliable parameter for diagnosing the bone-to-implant interface. Periometer ELC values are assumed to measure damping ratio but they did not correlate with the damping ratio values. The above analysis explains why the Periometer and damping ratio values did not agree.



[a]



[b]



[c]

**Figure 3.15** [a] Operational principle of the Perimeter. [b] Graph showing the energy return curve of the Perimeter. [c] Curve fitting of the graph to correct error in the Perimeter data.

#### 3.4.3.4 Fundamental issues with the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter.

From the discussion above, we can see that there are several fundamental issues behind the concept of non-invasive testing of dental implant stability. It is important to keep these

fundamental issues in mind. Otherwise, measurements from the Periotest<sup>®</sup>, the Osstell ISQ<sup>®</sup>, and the Perimeter could be easily misinterpreted or abused.

The first issue is that all these devices rest on the basic assumption of a single-degree-of-freedom system. For example, when a Periotest<sup>®</sup> or an Osstell ISQ<sup>®</sup> measurement is performed, a number (e.g., ISQ or PTV) will be obtained. Whether the tested implant along with its abutment or SmartPeg<sup>®</sup> satisfies the underlying assumption is not the question. If the assumption is not met, the number obtained is not accurate and cannot be used. Currently, there is no systematic way to check whether the single-degree-of-freedom assumption is met. This poses a real problem in using all these non-invasive testing devices, because it is not clear which measurements would be meaningful and which should be ignored.

The second issue is that all these devices do not directly measure implant stability. The Perimeter hypothesizes that damping correlates to implant stability. Therefore, damping properties are measured. Obviously, the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> hypothesize that the generalized stiffness  $k$  correlates to implant stability, i.e., the larger the stiffness, the better the stability. The generalized stiffness, however, may depend on many parameters, such as bone density, implant length, threads, and healing conditions at the implant-bone interface. To obtain the generalized stiffness  $k$ , the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> measure the natural frequency  $\omega_n$  and the contact time  $\tau$ , respectively. Extracting the generalized stiffness  $k$  accurately from the measured natural frequency  $\omega_n$  and contact time  $\tau$ , however, is not a trivial task. For example, the use of a SmartPeg<sup>®</sup> induces bending vibration of the implant. Therefore, the generalized mass  $m$  in equation (1) is not simply the mass of the implant. In fact, it is the inertia effect resulting from the entire bending vibration. It is not clear if algorithms embedded in the Osstell ISQ<sup>®</sup> and the

Periotest<sup>®</sup> convert the natural frequency  $\omega_n$  and the contact time  $\tau$  into generalized stiffness  $k$  with reasonable fidelity.

#### 3.4.4 *What is the true meaning of implant stability?*

To properly evaluate implant stability, one must answer the following question with no ambiguity: What physical quantity best represents implant stability? If osseointegration equates to implant stability, it is reasonable to hypothesize that stiffness  $k_{int}$  at the bone-to-implant interface best represents implant stability. If so, are the measurements from the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and EMA representative of implant stability?

Since FEA and EMA results are very consistent, we may use them to search for an answer. Here are some of our observations.

##### 3.4.4.1 Natural frequency values are not entirely representative of implant stability.

In the FEA study, the implant and Sawbones<sup>®</sup> blocks were perfectly bonded. The perfect bonding condition implies an infinite stiffness at the implant-Sawbones<sup>®</sup> interface. Furthermore, the density of the Sawbones<sup>®</sup> blocks does not affect the bonding condition at the interface. Therefore, the system has perfect implant stability under the hypothesis. The FEA results, however, showed that natural frequency values may change when the density of the Sawbones<sup>®</sup> blocks changes. Therefore, natural frequency does not entirely reflect the perfect bonding condition at the implant-Sawbones<sup>®</sup> interface.

The results above are not surprising, because natural frequency values are “system-level” quantities. In other words, natural frequency values are properties of the entire implant-bone-abutment system instead of properties at the bone-to-implant interface only (e.g., stiffness at the interface). In equation (1), the generalized stiffness  $k$  comes from any part of the system that

deforms, and the generalized mass  $m$  comes from any part of the system that moves with an inertial effect. According to the FEA results, the Sawbones<sup>®</sup> blocks deformed significantly, thus affecting the generalized stiffness  $k$  and subsequently the natural frequency  $\omega_n$ .

#### 3.4.4.2 The generalized stiffness $k$ extracted may not be representative of implant stability.

Since the natural frequency  $\omega_n$  is a system-level quantity, the generalized stiffness  $k$  derived from a measured  $\omega_n$  is also a system-level quantity. This feature manifested itself clearly in the FEA results. Figure 3.6 shows that the Sawbones<sup>®</sup> block and IMP abutment both deformed considerably to form the generalized stiffness  $k$ . If a generalized stiffness  $k$  is retrieved from a measured natural frequency  $\omega_n$  via EMA, it will reflect the stiffness of the IMP and the Sawbones<sup>®</sup> block rather than the implant-Sawbones interface.

How do we extract interfacial properties from system-level measurements? That is a major challenge in vibration engineering, and it is difficult to resolve. This will always be the case, regardless of which method is used to measure implant stability (even when we use a tensile machine to pull the implant).

The following example is an attempt to explain the nature of the problem. Assume the stiffness at the bone-to-implant interface is  $k_{int}$ , the stiffness of the implant-abutment assembly is  $k_{imp}$ , and the stiffness of the bone is  $k_b$ . When the implant-bone-abutment is tested, these three stiffness elements are in a series connection and a combined equivalent stiffness  $k_{eq}$  is measured. Since the elements are in a series connection, the equivalent stiffness is related to the individual stiffness coefficients via



$$\frac{1}{k_{eq}} = \frac{1}{k_{int}} + \frac{1}{k_{imp}} + \frac{1}{k_b} \quad (21)$$

If one stiffness coefficient among  $k_{int}$ ,  $k_{imp}$ , and  $k_b$  is significantly smaller than the other two, the measured stiffness  $k_{eq}$  will approximate the smallest stiffness. For example, if the interfacial stiffness  $k_{int}$  is significantly smaller than the bone stiffness  $k_b$  and the implant-abutment stiffness  $k_{imp}$ , then  $\frac{1}{k_b} \ll \frac{1}{k_{int}}$  and  $\frac{1}{k_{imp}} \ll \frac{1}{k_{int}}$ . Therefore, equation (21) becomes

$$\frac{1}{k_{eq}} \approx \frac{1}{k_{int}} \quad \text{or} \quad k_{eq} \approx k_{int} \quad (22)$$

When the stiffness of the implant-bone interface  $k_{int}$  is the same order as that of the implant-abutment stiffness  $k_{imp}$  or bone stiffness  $k_b$ , it is difficult to extract the interfacial stiffness  $k_{int}$  from the measured stiffness  $k_{eq}$ .

What are the implications of equations (21) and (22)? If a measured natural frequency  $\omega_n$  or generalized stiffness  $k$  is high, that means there is no weak spot. The interfacial stiffness  $k_{int}$  is most likely in par with other stiffness components. If a measured natural frequency  $\omega_n$  or generalized stiffness  $k$  is low, it may not necessarily indicate a weak interfacial stiffness  $k_{int}$ . It may result from low stiffness of the bone.

#### 3.4.5 *What are the advantages and limitations of experimental modal analysis?*

As mentioned previously, EMA is a common vibration test in mechanical engineering. Compared with the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter, EMA does not have the restriction of requiring a single-degree-of-freedom system. EMA is more versatile, measuring all sorts of quantities including natural frequency, damping ratio, and stiffness (if done correctly). Therefore, it allows us to investigate which parameter affects implant stability more prominently.

An effective approach in this study was the combined use of EMA and FEA. It is a highly effective strategy for two reasons. First, the results from FEA helped interpret the results obtained from EMA. Second, the combined use of FEA and EMA allowed us to identify sensitive parameters that needed to be well-controlled in our test models, thus substantially improving our experimental models. One such parameter identified in the study was residual stresses developed in the Sawbones<sup>®</sup> blocks. Residual stresses are permanent stresses left in an object after the pressure or loading applied to it is removed. Since the Sawbones<sup>®</sup> is made of polyurethane foam material, having it fixed repeatedly to a vise creates permanent stresses and affects the measurements. With the combined use of FEA and EMA, we can point out many problems encountered in the experiments. Based on our knowledge, the setup used in this study has not been previously applied on dental implants.

Nevertheless, EMA has its limitations. For example, it measures system-level parameters, not only those at the bone-to-implant interface. However, together with FEA it is a robust method to evaluate natural frequency of dental implants; it is very sensitive in identifying changes not detected by the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter. It can also serve as a benchmark to calibrate and evaluate dental implant stability devices.

### 3.5 Conclusions

Considering the limitations of the studies presented in this chapter, we conclude the following:

- The EMA measurements showed that dental implant length did not affect the natural frequency values but the simulated bone density did, results that were confirmed via FEA. EMA values were within 20% of FEA predictions.
- The Osstell ISQ<sup>®</sup> partially measured the frequency trend, while the Periotest<sup>®</sup> did not.
- The single-degree-of-freedom (SDOF) assumption used in the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> may limit their performance, as the tested implant-abutment bone systems are substantially more complex than an SDOF.
- The damping factor is a difficult parameter to quantify, potentially involving a significant degree of error.
- Major issues with the mechanical theories behind the Perimeter are reflected in the measurements obtained from this device.
- The use of Sawbones<sup>®</sup> as a study model should be exercised with caution. The relative size of the block, the uniformity of the block's material, number of implants placed in each block, and block fixation method could all affect the test results.
- EMA combined with FEA can serve as an effective benchmark for calibrating commercial dental testing devices.

The next chapter will observe how Sawbones<sup>®</sup> and porcine bones represent human mandibular bones. This is done by comparing the various types of bone models (human, animal,

and Sawbones<sup>®</sup>), by measuring with the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> PTV, and EMA natural frequency, while also assessing apparent bone density.

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## **CHAPTER 4. COMPARISON OF DIFFERENT TYPES OF BONE MODELS USED FOR PURPOSES OF DENTAL IMPLANT STABILITY TESTING *IN VITRO***

### **4.1 Introduction**

In section 2.5 of Chapter 2 (background and literature review), we described the different types of bone models used for purposes of dental implant stability testing *in vitro*. These include animal bones such as porcine bones, human cadaver jaw bones, and polyurethane blocks known as Sawbones<sup>®</sup>.

In the preceding chapter, we evaluated the use of both experimental modal analysis (EMA) and finite element analysis (FEA) in comparing results from the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter. The measurements obtained through both EMA and FEA demonstrated that implant length did not affect the natural frequency values but the simulated bone density did. In addition, the single-degree-of-freedom (SDOF) assumption used in the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> may limit their performance, as the tested implant-abutment bone systems are more



complex than a SDOF. Moreover, major issues with the mechanical theories behind the Perimeter were reflected in the measurements obtained from the Perimeter. The use of Sawbones<sup>®</sup> as a study model should be exercised with caution. The relative size of the block, uniformity of the block's material, number of implants placed in each block, and block fixation method could all influence test results. We concluded that the use of EMA, in combination with FEA, can serve as an effective benchmark for calibrating commercially available dental testing devices.

In this chapter, our aim is to observe how well Sawbones<sup>®</sup> and porcine bones represent human mandibular bones. Sawbones<sup>®</sup> are shaped to replicate the human and porcine bone models. We then compare the different types of bone models (i.e., porcine bone, human mandibular bone, and Sawbones<sup>®</sup>), using the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and EMA. Moreover, FEA for the Sawbones<sup>®</sup> porcine bone replica is carried out to verify the measurements. In addition, the two Sawbones<sup>®</sup> models (human and porcine bone replicas) are compared. Apparent density values are calculated for each bone sample and then correlated with EMA, the Osstell ISQ<sup>®</sup>, and the Periotest<sup>®</sup> measurements.

## **4.2 Materials and Methods**

### *4.2.1 Bone samples.*

Dry bone samples were used in this study for comparison with Sawbones<sup>®</sup>. Dry bone loses the viscoelastic properties present in fresh bone. Fresh bone should be frozen and then thawed before being used in an experiment. This process might affect the mechanical properties of the bone (Lee & Jasiuk, 2014). Because of complications relating to the use of fresh bone, for the

purpose of using samples similar to Sawbones<sup>®</sup>, dry bone was used in this study. Three different types of bone models were used: porcine bones, human bones, and Sawbones<sup>®</sup>.

Two samples of porcine femur (from pigs 12-18 months of age) were obtained from Dr. Susan Herring's laboratory (Oral Health Sciences, School of Dentistry, University of Washington, Seattle, USA). The samples measured an average of 39mm in (mediolateral mid-shaft) width, a full cross-sectional segment of 21 mm in length, 6.5 mm in thickness, and an intramedullary canal diameter of 11.5 mm. The cross-section of the samples can be seen in Figure 4.1.

Two samples of human mandibular bones (premolar-molar region) of unknown age—measuring an average of 30 mm in width, a full cross-sectional segment of 26mm in length, and a cross sectional width of 15mm—were obtained from the School of Medicine's Anatomy Department, University of Washington, Seattle, USA. Cross-sections of the models are pictured in Figure 4.1.

40-PCF ( $0.64 \text{ g/cm}^3$ ) Sawbones<sup>®</sup> blocks were used. The blocks were modeled to match the shape and dimensions of the porcine and human bone samples, cross-sections of the models are pictured in Figure 4.1.



**Figure 4.1** Cross sectional/side views of the different bone models used in the study.

#### 4.2.2 *Dental implants.*

Branemark Mk III implants measuring 4 mm in diameter and 7 mm in length were placed in the center of each bone sample. All implants were placed following the manufacturer's surgical protocol, and seated to 45 N·cm.

#### 4.2.3 *Boundary conditions.*

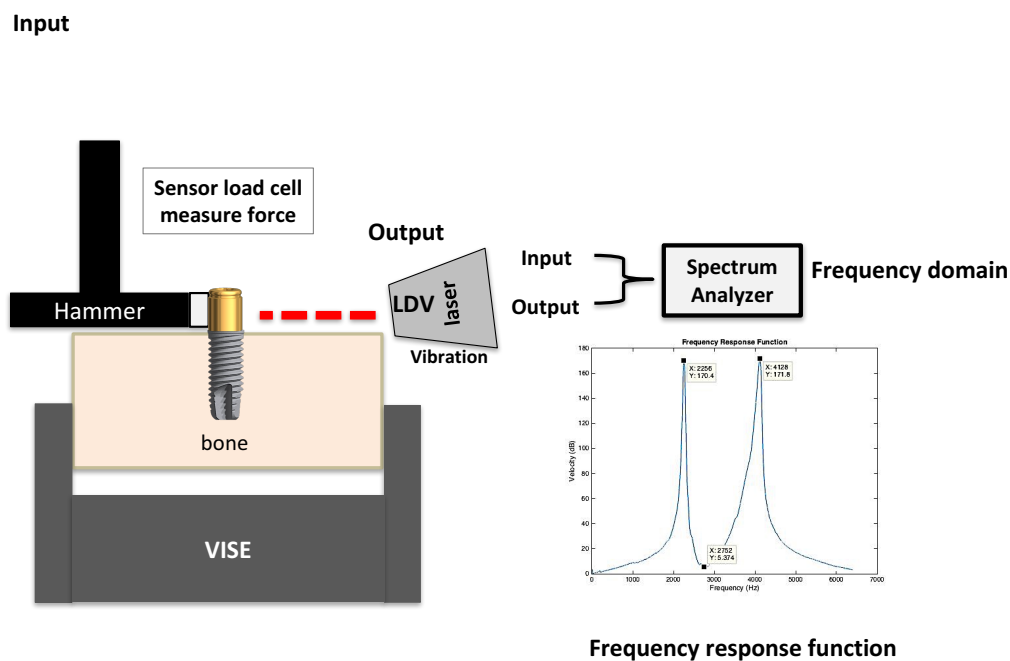
All samples were secured with a lateral anchorage, where the samples were fixed on both sides. Approximately the lower half of each sample was firmly clamped by a vise bolted to an isolation table, so as to reject possible external disturbances. Two parallel sides of the samples were therefore in a fixed boundary condition.

#### 4.2.4 *Measurement methods.*

The implant-bone systems were measured using the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and experimental modal analysis (EMA). For the Osstell ISQ<sup>®</sup>, a SmartPeg<sup>®</sup> (type I) was screwed on the implant at the fixture level with finger pressure. Mesial (a) and distal (b) measurements were repeated three times, after which averages were recorded. For the Periotest<sup>®</sup> and EMA, a locator abutment (NobelBiocare, Switzerland) (LA) 7 mm in length was used. A 10 N·cm torque was used to tighten the abutment with Branemark system torque control. For the Periotest<sup>®</sup> measurements, the samples were tested a few millimeters away, with the device tip placed perpendicular to the access of the implant.

For all samples, mesial (a) and distal (b) measurements were repeated three times and an average was recorded.

The EMA setup described in the previous chapter consisted basically of a force hammer, a laser Doppler vibrometer (LDV), and a spectrum analyzer. The hammer measured the force applied to the implant, while the LDV measured the velocity response of the implant (see Figure 4.2). Based on the measured force and velocity, the spectrum analyzer calculated frequency response functions, from which the first natural frequency values of the test models were extracted. Mesial (a) and distal (b) natural frequency was measured at three different times after which the mean was recorded.



**Figure 4.2** Experimental modal analysis (EMA) setup. Approximately the lower half of each sample was firmly clamped by a vise. The samples were fixed on both sides.

#### 4.2.5 *Finite element analysis (FEA).*

A 3D finite element model was created using ANSYS Workbench 17.2 (Canonsburg, PA) to simulate the experimental setup and the test results of porcine bone replica. The finite element model (FEM) consists of three parts: A Sawbones<sup>®</sup> model, a cylindrical implant, and a LA. The Sawbones<sup>®</sup> block was modeled as a 39 mm hollow cylinder with similar dimensions to the porcine femur. The material properties for the 40-PCF were used to model the hollow cylinder (see Table 3.1). The Sawbones<sup>®</sup> model was assumed to be isotropic. Moreover, the ends of the bone block were fixed using a fixed support condition up to the lower half of the sample height, similar to the boundary conditions followed in the experiments. To model the implant, a one-piece titanium cylinder, 7 mm in length and 4 mm in diameter, was used. Moreover, the cylindrical implant was located at the center of the Sawbones<sup>®</sup> model. The nodes of the cylindrical implant and the Sawbones<sup>®</sup> block were merged at the implant-Sawbones<sup>®</sup> interface to model a no-slip condition (i.e., perfect bonding) between the implant and the Sawbones<sup>®</sup> block. The LA abutment with approximately the same dimensions was modeled as a titanium cylinder and was then connected to the implant via a no-slip condition. An auto-generated fine mesh was used with a 0.75 mm maximum edge length on the bone cylinder. After the model was created, a modal analysis was conducted to predict the first natural frequency of the simulated test sample.

#### 4.2.6 *Apparent bone density estimation.*

10 sample cubes measuring approximately (0.5 X 0.5 X 0.5) cm were prepared from each bone sample and Sawbones<sup>®</sup> blocks. The apparent bone density (using bone samples consisting of 90% trabecular bone) was estimated by dividing the mass (m) of the cube (grams) by the volume (V) of the cube in cm<sup>3</sup>, using the equation:

$$\text{Apparent Density} = \frac{m}{V}$$

#### 4.2.7 *Statistics.*

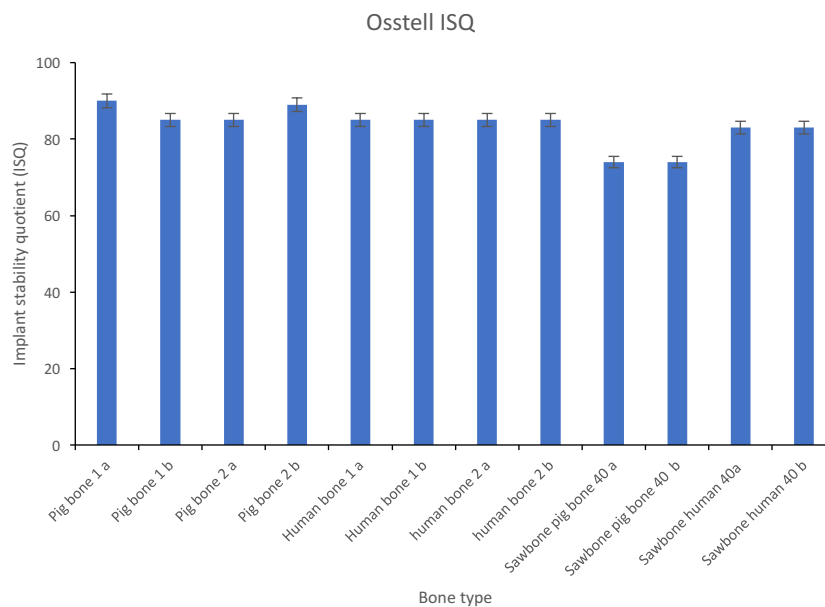
The mean ISQ, PTV, natural frequency values, and the apparent density were calculated, as well as their standard deviations (SD). The correlation coefficient between apparent bone density and these values (ISQ, PTV, and natural frequency values) was estimated.

## 4.3 Results

### 4.3.1 *Porcine femurs vs human mandibular bones.*

#### 4.3.1.1 Osstell ISQ®.

The mean ISQ values for the two porcine bone samples fell within the same range (85-90). In addition, the ISQ values for two human mandibular bone samples also fell within the same range (85). The ISQ values for the porcine bone and mandibular bone samples did not differ significantly as seen in Figure 4.3.



**Figure 4.3** The mean Osstell ISQ® values for the various types of bone models with measurements taken mesially (a) and distally (b).

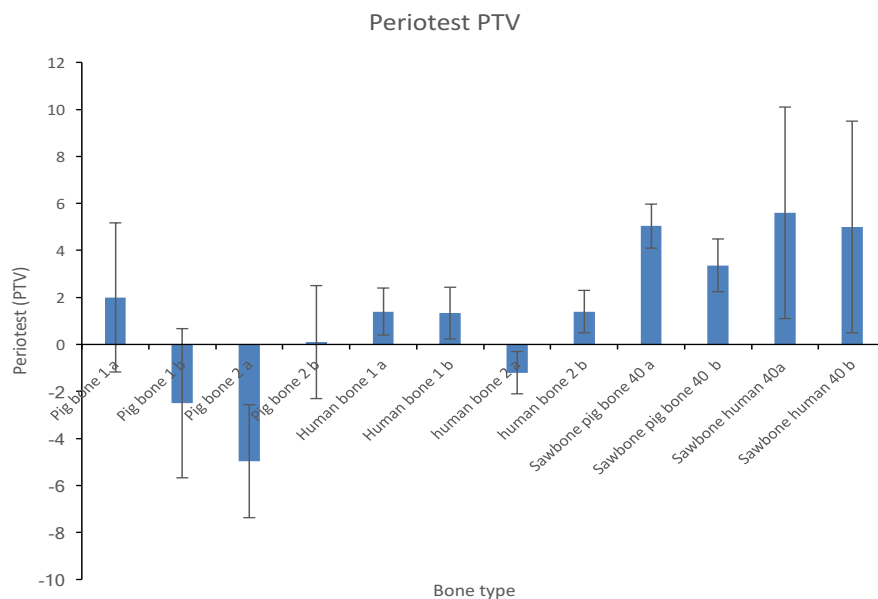
#### 4.3.1.2 Periotest PTV®.

In general, the PTV readings were inconsistent with a high SD. For both porcine bone samples, a significant difference is seen in the mean PTV readings for sides (a) and (b), as observed



in Figure 4.4. The human mandibular bone samples, however, did not differ significantly, except human mandibular bone 2 side (a) had a significant lower value, as shown in Figure 4.4.

When comparing human and porcine bone samples, it appears that the porcine bone samples in general have significantly lower PTV readings than human bone samples, except for porcine bone sample 1 side (a) which has a higher PTV value (see Figure 4.4).

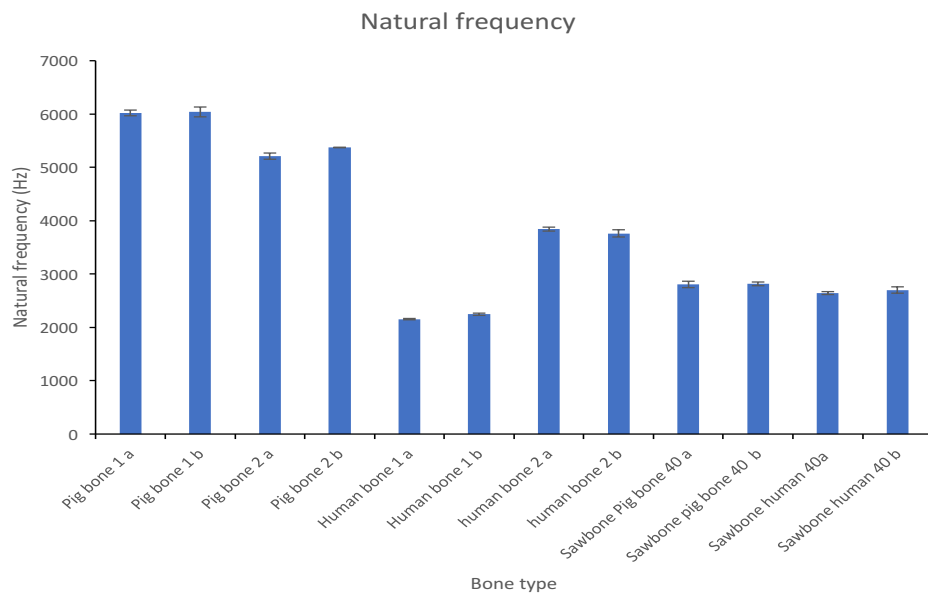


**Figure 4.4** The mean PTV for the various types of bone models with measurements taken mesially (a) and distally (b).

#### 4.3.1.3 Experimental modal analysis (EMA) natural frequency values.

The average natural frequency values for porcine bone 1 side (a) was 6021.3 Hz and side (b) was 6040 Hz. The values for porcine bone 2 on the other hand, were slightly lower, 5211 Hz for side (a) and 5373.3 Hz for side (b).

The average natural frequency values for the human mandible 1 were 2154 Hz for side (a) and 2246 Hz for side (b). While the frequency values for human mandible 2 were slightly higher, 3842.7 Hz for side (a) and 3762.7 Hz for side (b) (see Figure 4.5). The frequency values for porcine femur samples were significantly higher than the frequency values for the human mandibular bone samples.



**Figure 4.5** The mean experimental natural frequency values for the different bone models with measurements taken mesially (a) and distally (b).

#### 4.3.1.4 Apparent density values.

For the porcine bone samples the apparent density values were around 0.9-1.6 g/cm<sup>3</sup> (SD  $\pm$  0.256) for sample 1 and 1.2-1.9 g/cm<sup>3</sup> (SD  $\pm$  0.26) for sample 2. The human mandibular bone

samples had slightly lower values. The apparent density values ranged from 0.8-1.1 g/cm<sup>3</sup> (SD  $\pm$  0.087) for sample 1, and between 0.7-1.4 g/cm<sup>3</sup> (SD  $\pm$  0.236) for sample 2 (see Table 4.1).

**Table 4.1** Estimated apparent bone density values for all bone samples.

| APPARENT BONE DENSITY (g/cm <sup>3</sup> ) |              |              |             |            |            |
|--------------------------------------------|--------------|--------------|-------------|------------|------------|
| Sample number                              | Human bone 1 | Human bone 2 | Sawbones 40 | Pig bone 1 | Pig bone 2 |
| 1                                          | 1.1          | 1.3          | 0.629       | 1.25       | 1.54       |
| 2                                          | 0.87         | 0.99         | 0.57        | 1.2        | 1.8        |
| 3                                          | 0.91         | 0.87         | 0.634       | 1.6        | 1.23       |
| 4                                          | 0.87         | 1.23         | 0.636       | 1.56       | 1.82       |
| 5                                          | 0.80         | 0.68         | 0.65        | 1.71       | 1.9        |
| 6                                          | 1.027        | 0.81         | 0.46        | 1.35       | 1.2        |
| 7                                          | 0.94         | 1.38         | 0.88        | 0.93       | 1.32       |
| 8                                          | 0.95         | 0.8          | 0.66        | 0.99       | 1.56       |
| 9                                          | 0.853        | 0.97         | 0.86        | 1.24       | 1.4        |
| 10                                         | 0.91         | 0.80         | 0.62        | 1.46       | 1.3        |
| AVERAGE                                    | 0.923        | 0.983        | 0.66        | 1.33       | 1.51       |
| SD                                         | 0.087        | 0.236        | 0.124       | 0.256      | 0.26       |

#### 4.3.2 *Porcine bone vs Sawbones<sup>®</sup>*.

##### 4.3.2.1 Osstell ISQ<sup>®</sup>.

The mean ISQ values for the Sawbones<sup>®</sup> porcine bone replica was 74 for both sides. These values are significantly lower than the ISQ for porcine bones, see Figure 4.3.

##### 4.3.2.2 Periotest PTV<sup>®</sup>.

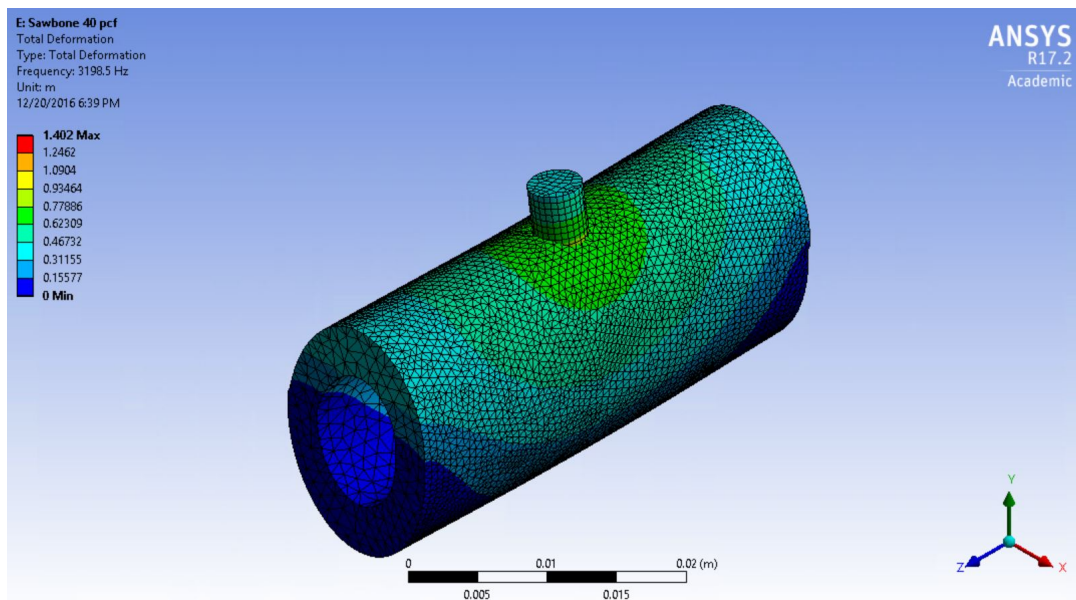
In general, the mean PTV for the Sawbones<sup>®</sup> porcine bone replica was significantly higher than the PTV values for the porcine bones (see Figure 4.4).

#### 4.3.2.3 EMA natural frequency measurement.

The average natural frequency values for the Sawbones<sup>®</sup> porcine bone replica was 2805.3 Hz for side (a) and 2816 Hz for side (b), as demonstrated in Figure 4.5. The natural frequency values of the Sawbones<sup>®</sup> porcine bone replica were significantly lower than for the porcine bones values.

#### 4.3.2.4 FEA.

The model for the first mode can be seen in Figure 4.6. The first natural frequency prediction for Sawbones<sup>®</sup> porcine bone replica was 3198.5 Hz. The EMA results are about 14-16 % of the FEA value.



**Figure 4.6** Finite element analysis (FEA) of the Sawbones<sup>®</sup> porcine femur replica, (first natural frequency mode) created using ANSYS.

#### 4.3.2.5 Apparent bone density values.

The apparent density values for the Sawbones<sup>®</sup> were significantly lower than that of the porcine bones. For Sawbones<sup>®</sup>, the apparent density ranged from 0.46 to 0.88 g/cm<sup>3</sup> (SD  $\pm$  124).

#### 4.3.3 *Human mandibular bones vs Sawbones<sup>®</sup>*

##### 4.3.3.1 Osstell ISQ<sup>®</sup>.

The ISQ values for the Sawbones<sup>®</sup> human bone replica and the human mandibular bones did not differ significantly, as illustrated by Figure 4.3.

##### 4.3.3.2 Periotest PTV<sup>®</sup>.

The mean PTV for the Sawbones<sup>®</sup> human bone replica was significantly higher than the PTV values for the human mandibular bones (see Figure 4.4).

##### 4.3.3.3 EMA natural frequency values.

The average frequency values for the Sawbones<sup>®</sup> human bone replica were 2643.2 Hz for side (a) and 2700.8 Hz for side (b). This is closer to human mandible sample 1 values and significantly lower than the values for human mandible sample 2, as seen in Figure 4.5.

##### 4.3.3.4 Apparent density values.

The apparent density values for the Sawbones<sup>®</sup> were close to the lower range of the estimated apparent density for human mandibular bone samples (see Table 4.1).

A positive correlation was seen between the natural frequency values and the apparent density values for all bone samples, indicating that the natural frequency values increase with

increases in apparent density. No correlation was observed between the apparent bone density for all samples and the Osstell ISQ<sup>®</sup> and PTV readings.

#### 4.3.4 *Sawbones<sup>®</sup> models differing in geometry.*

To see whether the geometry of the models affected the measurements, we separately compared the readings obtained from the Sawbones<sup>®</sup> human and porcine bone replicas, as described immediately below.

##### 4.3.4.1 Osstell ISQ<sup>®</sup>.

The ISQ values obtained from the human replica were significantly higher than for the porcine bone replica, as shown in Figure 4.3.

##### 4.3.4.2 Periotest<sup>®</sup> PTV.

The PTV readings for the Sawbones<sup>®</sup> human bone replica were much higher than the PTV values for the porcine bone replica (see Figure 4.4).

##### 4.3.4.3 EMA natural frequency values.

The average natural frequency values for the human mandible replica were around 150 Hz lower than the values for the porcine bone replica, as illustrated in Figure 4.5.

## 4.4 Discussion

The objective of this chapter, was to assess how well Sawbones<sup>®</sup> and porcine bones represent human mandibular bones. Sawbones<sup>®</sup> were shaped to replicate the human and porcine bone models, after which we compared the different types of bone models (i.e., porcine bone,

human mandibular bone, and Sawbones<sup>®</sup>), using the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and EMA, while assessing apparent bone density. FEA for the Sawbones<sup>®</sup> porcine bone replica was carried out to verify the measurements. Comparing the two Sawbones<sup>®</sup> models (human and porcine bone replicas), apparent density values were calculated for each bone sample and then correlated with the EMA, the Osstell ISQ<sup>®</sup>, and the Periotest<sup>®</sup> measurements.

For the samples to be compared on an equal footing, the following were done: First, identical implants were placed into the different bone types, using a similar surgical technique, using both dry human and porcine bone samples. Sawbones<sup>®</sup> were then shaped to replicate the geometry of the porcine and human mandibular bone samples. Finally, all samples were tested under similar fixed boundary conditions.

It should be noted that this study was done for purposes of comparing the different bone samples, rather than for the purpose of evaluating implant stability. As concluded in the preceding chapter, natural frequency does not represent implant stability. Natural frequency was used in this study because it represents a robust measurement, and we have seen how it changes significantly with changes in the density of the material.

The natural frequency values differed between the bone samples. Although obtained from the same animal, the natural frequency values were greater for porcine bone sample 1 than for porcine bone sample 2. This illustrates how bone can be a challenging material to work with and the results may be difficult to reproduce. Although the same boundary conditions were carefully used, any slight variation in the position or angulation of the sample clamped by the vise might affect the results. Therefore, sensitivity to the experimental setup should be taken into consideration, evaluated conscientiously. The same issue presented itself with the mandibular

samples. The natural frequency values were higher for mandibular bone sample 2 than for sample 1, illustrating again how actual bones present unique mechanical and physical properties. Also, examining the cross section of the human mandibular bone samples in Figure 4.1, sample 1 shows a more porous trabecular network than sample 2. This might have played a role in the lower apparent density and natural frequency values evidenced in human mandibular bone sample 1.

Although neither the Osstell ISQ<sup>®</sup> nor the Periotest<sup>®</sup> provide an accurate estimation of implant stability, as discussed in the preceding chapters, both devices are nonetheless used in many studies to measure implant stability, in part because no better option is available and in part probably because the researchers are not aware of the devices' limitations (Al-Nawas et al., 2006; Chan et al., 2010; Degidi et al., 2017; Ersanli et al., 2005; Friberg et al., 1999; Krafft et al., 2015; Ostman et al., 2006). In the present study, the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> are compared with regard to their ability to measure natural frequency values.

Apparent density has been defined as “mass of bone tissue[,] divided by the bulk volume of the test specimen, including mineralized bone and marrow space” (Bouxsein & Hayes, 1997). Trabecular bone is a porous structure, with numerous large spaces filled with bone marrow that becomes brittle when dried, and a larger surface-to-mass ratio than the cortical bone, which is a fairly uniform mineralized solid matrix (Seong et al., 2009a; Seong et al., 2009b). Composed of a combination of irregular columns and struts, trabecular bone's column formation is random, resulting in hollow spaces of different diameters. The density of the human mandible varies greatly among individuals. It also varies according to age, gender, and the presence or absence of teeth, as well as in different parts of the mandible. The anterior part of the mandible may have more cortical bone density compared to the posterior portion (Bryant, 1998; Ko et al., 2017; Turkyilmaz et al.,



2009). Cortical bone tends to be more dense than trabecular bone, also known as cancellous bone. Our samples were taken from dry old skull mandibles, obtained from the School of Medicine's Anatomy Department at the University of Washington, Seattle, WA. The tested density samples consisted primarily of trabecular bone, with a small percentage of cortical (approximately 10%).

The apparent density for human mandibular bone measured in this study ( $0.8\text{-}1.4\text{ g/cm}^3$ ) is similar to what has been reported in other studies. Our results agree with those of O'Mahony et al. (2001) who estimated the trabecular bone density to be within the range of  $0.9\text{-}1.2\text{ g/cm}^3$ . Seong et al. (2009b) estimated mandibular bone density to be  $0.96\text{-}1.18\text{ g/cm}^3$ .

The apparent density of this study's porcine femur bones ( $0.9\text{-}1.9\text{ g/cm}^3$ ) is slightly lower than what was reported by Lee and Jasiuk (2014), who estimated the density to be  $2.27\text{ g/cm}^3$ . Such data are hard to find in the literature.

The apparent density value for 40-PCF Sawbones<sup>®</sup> as reported by the manufacturer is  $0.64\text{ g/cm}^3$ . When measured in our study, the apparent density for Sawbones<sup>®</sup> ranged from  $0.46\text{ to }0.88\text{ g/cm}^3$ , within the  $\pm 10\%$  precision described by Sawbones<sup>®</sup>.

#### 4.4.1 *Porcine bones vs. human mandibular bones.*

Bones from pigs are typically considered desirable for experimental studies relating to humans, as porcine bone structure is considered very close to that of humans (Pearce et al., 2007). But how well do porcine bones in fact represent human mandibular bone, for purposes of dental implant studies? As seen in our results, porcine bone has higher apparent density values than do human bones, which significantly affects natural frequency values. Porcine bone also showed significantly higher natural frequency values. The Periotest PTV<sup>®</sup> readings in general were very inconsistent and inconclusive, with a high SD, even for the different sides of the same sample.

These findings suggest lack of reliability with regard to the ability on the part of the Periotest<sup>®</sup> to measure bone density. The Osstell ISQ<sup>®</sup> values did not differ much between the samples, which again suggests lack of accuracy on the part of the Osstell ISQ<sup>®</sup> device for comparing implants placed in different bone densities.

From the limitations reflected in this study, it would be difficult to conclude that porcine bones do a good job of representing human bones. Porcine bones are of much higher density than human mandibular bones. As porcine bones may also generate exaggerated results in cases in which the stability of the implant is estimated, any conclusions based on such data should therefore be considered with caution.

#### 4.4.2 *Porcine bones vs Sawbones<sup>®</sup>.*

40-PCF Sawbones<sup>®</sup> were modeled to replicate samples of porcine bones, the apparent density values of which appear to be much higher than that of the Sawbones<sup>®</sup>. Also, the natural frequency values obtained from porcine bones in this study were significantly greater was the case for the Sawbones<sup>®</sup> values (see Figure 4.5). The Osstell ISQ<sup>®</sup> values were slightly lower for the Sawbones<sup>®</sup>, in contrast with the PTV, which was higher for the Sawbones<sup>®</sup> than the porcine bone samples. Comparing Sawbones<sup>®</sup> to porcine bones is somewhat problematic, however, as the two types vary significantly in apparent density values.

In our previous study, FEA offered an extreme prediction of the EMA results. Because we were unable in this study to perform an FEA on either porcine or human bone, we created a finite element model for the Sawbones<sup>®</sup> porcine bone replica, in order to determine the accuracy of the natural frequency readings. The FEA results confirmed the first natural frequency values for Sawbones<sup>®</sup> measured with EMA, with an FEA value within 14 % of the EMA results.

#### 4.4.3 *Human bones vs Sawbones®.*

As discussed above, Sawbones® are used prominently for purposes of *in vitro* dental implant studies, to represent human jaw bones, as the Sawbones® offer a model that is much cleaner and easier to use. In this study, the natural frequency values of the Sawbones® were slightly lower but closer to the values of human mandible sample 1. Human mandible sample 2 values were much higher, however. The apparent density values for the Sawbones® 0.4-0.88 g/cm<sup>3</sup> approach the apparent density range of 0.87-1.1 g/cm<sup>3</sup> that was estimated for the human mandible bone models.

Although Sawbones® may constitute a good model for representing human bone, any resulting data should be used with caution, as Sawbones® do not accurately reflect actual bone biomechanics. Sawbones® are usually used for the purpose of obtaining comparable biomechanical results, rather than definitive, clinically applicable, information.

#### 4.4.4 *Sawbones® models differing in geometry.*

Sawbones® of identical density (40-PCF) but different geometry were compared to observe the effect of the geometry on the measurements. The porcine bone replica was slightly larger in terms of dimensions, relative to the human bone replica. Geometry or shape of bone might affect the EMA readings. Although the natural frequency values were slightly higher for the Sawbones® porcine bone replica, the Osstell ISQ® and PTV values for the Sawbones® human mandibular bone replica were higher than the values for the porcine bone replica.

Comparing our results with those of previous studies is problematic, as many studies place multiple implants in a single bone or Sawbones® sample (Bilhan et al., 2015; Chan et al., 2010; Dalton et al., 2003; Dard et al., 2016; de Oliveira et al., 2016; Wang et al., 2015; Yang et al., 2002),

Because we found that such placement of multiple implants complicates the readings and do not offer an accurate representation, we did not follow this practice. The standard use of fresh frozen bones in studies makes it difficult to compare results generated by the use of dry bone values (Degidi et al., 2017; Kheur et al., 2016; Sivoilella et al., 2015). In many tests that use animal bones, furthermore, bone density is poorly defined, often based on assumptions rather than actual measurements (Andres-Garcia et al., 2009; Bilhan et al., 2015; Degidi et al., 2017; Kheur et al., 2016).

## **4.5 Conclusions**

Given the limitations inherent in this study, as well as the limitations uncovered by this study, in this chapter we conclude the following:

- Porcine bone models evidenced both higher apparent density values and much higher frequency values than human bone models, suggesting that porcine bone models might overestimate implant stability measurements.
- The Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV were unable to differentiate implants in various bone models, suggesting their unreliability for this purpose.
- Although Sawbones<sup>®</sup> may overall offer a good model for representing human bone, measurements resulting therefrom should be considered with caution, as Sawbones<sup>®</sup> do not accurately reflect actual bone biomechanics. Sawbones<sup>®</sup> are usually selected for achieving comparable biomechanical results, rather than definitive, clinically applicable, information.

The chapter that follows will utilize micro-CT to examine bone mineral density and implant-bone morphometrics in human and porcine bone models.

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## **CHAPTER 5. BONE MINERAL DENSITY ESTIMATION AND MORPHOMETRIC ANALYSIS OF THE BONE-TO-IMPLANT INTERFACE, USING MICRO- COMPUTED TOMOGRAPHY (MICRO-CT) *IN VITRO***

### **5.1 Introduction**

The experimental methods used to measure dental implant stability were discussed in section 2.4.2 of Chapter 2. Histomorphometric analysis, as discussed above, is the standard type of measurement used in experimental studies to calculate the amount of bone in contact with an implant surface (BIC). Histomorphometric analysis, however, has the distinct disadvantage of being a destructive method that leads to adverse changes or damage to the tissues and implants during preparation. It was also mentioned that micro-CT is a non-destructive method for obtaining morphometric measurements of bone and peri-implant area, i.e., BIC. Micro-CT is also used to estimate the bone mineral density (BMD) of scanned samples.



In the preceding chapter, we observed how Sawbones<sup>®</sup> and porcine bones were used in the previous study to represent human mandibular bones. We compared the different types of bone models (i.e., porcine bone, human mandibular bone, and Sawbones<sup>®</sup>), using the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> devices, and EMA, in addition to measuring apparent bone density. We concluded that porcine bone models might result in overestimating implant stability. The Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV were unable to differentiate between implants in different types of bone models, suggesting their unreliability for this purpose. Although Sawbones<sup>®</sup> may overall offer a good model for representing human bone, measurements resulting therefrom should be considered with caution, as Sawbones<sup>®</sup> do not accurately reflect actual bone biomechanics. Sawbones<sup>®</sup> are usually selected for achieving comparable biomechanical results, rather than for obtaining definitive, clinically applicable, information.

In this chapter, we further examine the human mandible and porcine femur bone models, using micro-CT imaging, in order to: (1) estimate the BMD and (2) perform a morphometric analysis of bone and the bone-to-implant interface.

## **5.2 Materials and Methods**

### *5.2.1 Bone samples.*

Porcine bone femur and human mandibular bone samples described in chapter 4, section 4.2.1, were used in this portion of the study.

### 5.2.2 *Dental implants.*

As described in the preceding chapter, Branemark Mk III implants (4 mm in diameter and 7 mm in length) were placed in the center of each sample. All implants were placed according to the manufacturer's surgical protocol seating, to 45 N·cm.

### 5.2.3 *Micro-CT scans.*

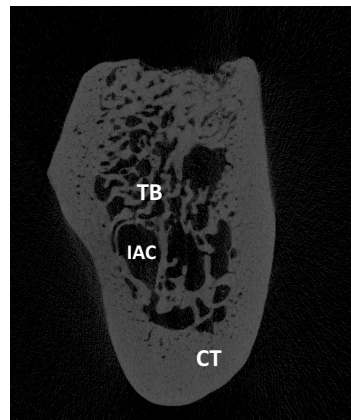
All scans were performed using the Skyscan 1076 desktop X-ray Micro-tomography System (Skyscan N.V., Belgium) at the Small Animal Tomographic Analysis Facility (SANTA) at the University of Washington's Department of Pediatrics, School of Medicine, Seattle, WA. The micro-CT scans were used for estimating BMD and morphometric analysis of the bone-to-implant interface, as described below.

#### 5.2.3.1 Estimating BMD.

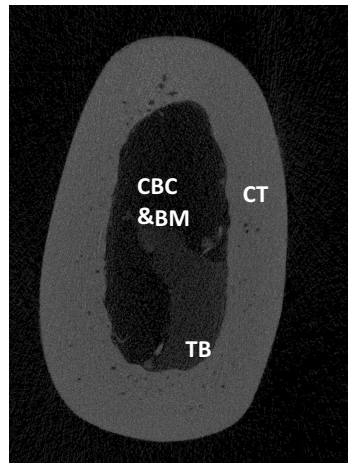
The bone samples were scanned at an energy of 104 kV and an intensity of 96  $\mu$ A, with a pixel resolution of 35 $\mu$ m and 250 m.s exposure. The protocol in Bruker-MicroCT method note for estimating BMD was followed (Bruker-MicroCT, 2005). Bruker-MicroCT BMD phantoms (rat tibia) of known density were used for calibration of the bone samples. The calibration phantoms come in pairs, with calcium hydroxyapatite (CaHA) concentrations of 0.25 and 0.75 g/cm<sup>3</sup>. The images were then re-constructed using NRecon software (version 1.6.9.4; Skyscan, Bruker, Belgium) after applying the ring artifact correction and beam hardening correction tools (set at 8% and 40%, respectively), as pictured in Figure 5.1.

The BMD was calculated using CT-Analyser (CTAn) software (Skyscan, Bruker, Belgium). After the BMD of the phantoms was determined, the attenuation coefficients for the

phantoms were used to calibrate the bone samples. The BMD of selected images of the premolar and molar regions for the human mandibles was calculated, as was done for regions of the porcine bone femur. Because of the density difference between the bone samples (both human and porcine) and the phantom rat samples used for calibration, a 20% precision was expected in the estimated BMD.



[a] Human mandibular bone



[b] Pig femur

CT: Cortical bone  
TB: Trabecular bone  
IAC: Inferior alveolar canal  
CBC : Central bone cavity  
and  
BM: Bone marrow space

**Figure 5.1** Micro-CT cross-section NRecon images used to estimate BMD for [a] human mandibular bone and [b] porcine bone femur.

#### 5.2.3.2 Morphometric analysis of the bone-to-implant interface.

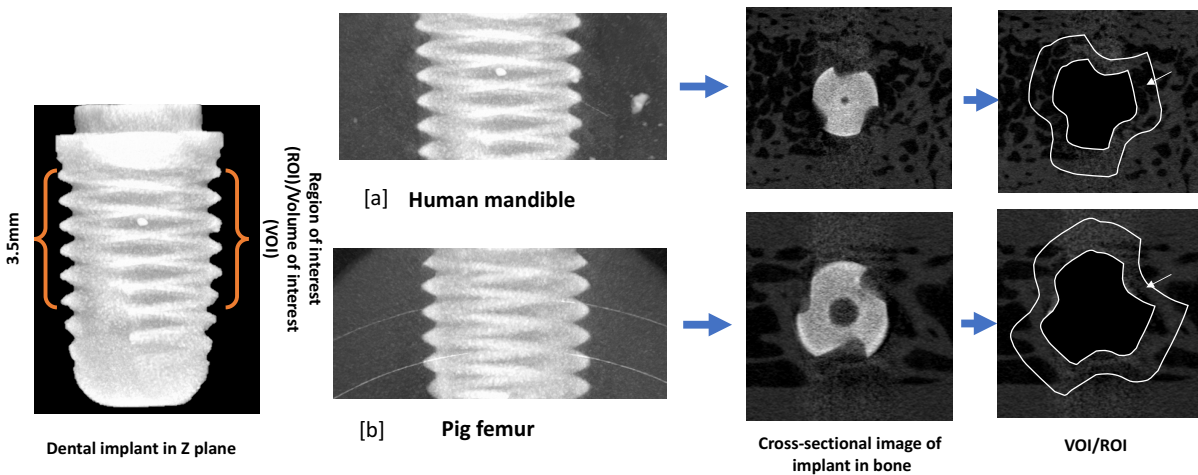
A 3-cm area around the implant was scanned in each bone sample at an energy of 59 kV and an intensity of 149  $\mu$ A, with a pixel resolution of 18  $\mu$ m. A 0.5-mm-thick aluminum filter was used, at a 180° rotation, with a rotation step of 0.7°, at 400 m.s exposure. The images were then re-constructed using NRecon software after applying ring artifact correction and beam-hardening correction tools (set at 8% and 40%, respectively).

Bruker-MicroCT method note for analysis of bone around a metal implant was followed (Bruker-MicroCT, 2015). First, the dataset was adjusted in Data Viewer (SkyScan, Bruker, Belgium), so the implant's long axis was orthogonal to the cross section's XY plane. Then we saved a new dataset in which the images were realigned and restricted to only the necessary volume around the implant. The data were then analyzed using the CTAn software, to calculate the morphometric parameters. To compare the samples, the same volume of interest (VOI) (for 3D analysis) and region of interest (ROI) (for 2D analysis) were selected for all samples. The VOI of each sample was calculated by determining the region of interest (ROI) in each cross-sectional image acquired from the whole volume sample. To standardize the vertical range of the VOI for all samples, a fixed number of slices was chosen for both the offset (distance between the red reference level and the start of the selected range) and the height of the VOI (number of cross-section levels from top to bottom of the selected range). These two values—the offset and the height—together with the location of the reference level, collectively define the vertical selected range. The VOI/ROI selected was between the 1<sup>st</sup> and 6<sup>th</sup> implant threads embedded in bone, approximately 3.5 mm in height, as pictured in Figure 5.2. Bone morphometric parameters were calculated for each sample, following the Bruker-MicroCT method note (Bruker-MicroCT, 2015).

The upper and lower threshold levels for bone and implants were determined in all samples and did not overlap, allowing for a clear distinction. Threshold values were set for bone (max. 100, min. 43) and dental implants (max. 255 and min. 110). Because of the artifacts caused by the titanium implant on the micro-CT scans, the first 10 pixels from the implant surface were excluded, for purposes of unbiased morphometric analysis. Subsequent analysis of the scans was performed using the 3D data analysis in CTAn. The following morphometric parameters were obtained using 3D analysis from micro-CT scanning, which offers an intensive analysis of peri-implant bone area:

- The total volume (TV) of the entire region of interest, measured in  $\text{mm}^3$ .
- The bone volume (BV) of the region segmented as bone (or mineralized) tissue, measured in  $\text{mm}^3$ .
- The bone surface (BS) region, segmented as bone, measured in  $\text{mm}^2$ .
- The intersection surface (i.S) area of implant in contact with bone, measured in  $\text{mm}^2$ .
- The bone volume fraction (BV/TV) ratio of the bone volume to the total volume of the region, a parameter described as a percentage, used to represent BIC.
- The bone surface density (BS/TV) ratio of the bone surface to the total volume, measured in  $\text{mm}^2/\text{mm}^3$ .
- The specific bone surface (BS/BV) ratio of the bone surface to the bone volume, measured in  $\text{mm}^2/\text{mm}^3$ .
- The (mean) trabecular thickness (Tb.Th), calculated from the trabecular BV and TV measured in mm.
- The trabecular number (Tb.N), a measure of the average trabeculae per unit length, measured in 1/mm units.

- The trabecular separation (Tb.Sp), the distance between adjacent trabeculae, measured in mm.



**Figure 5.2** Volume of interest (VOI)/region of interest (ROI) for each sample used for the micro-CT analysis. The area of the implant between the 1<sup>st</sup> and 6<sup>th</sup> thread was used to define VOI /ROI. The morphometric parameters were estimated from the VOI/ROI.

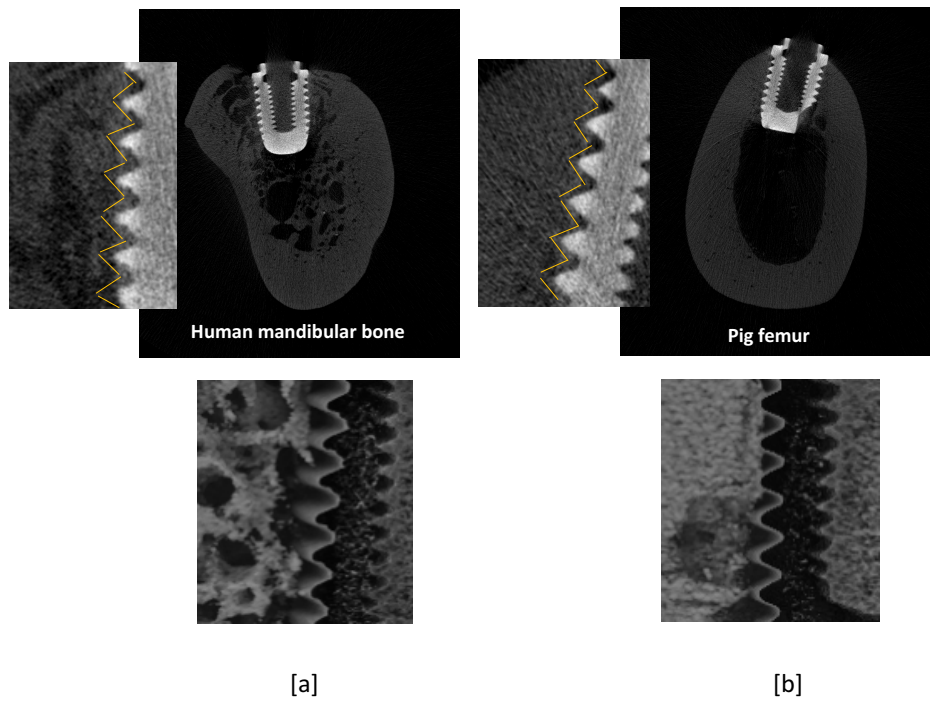
The micro-CT 2D analysis is often used to estimate i.S./TS, the bone-to-implant intersection surface (measuring the percent of the implant surface in contact with nearby bone, known as BIC).

#### 5.2.3.3 Measuring BIC manually from micro-CT images.

As seen in Figure 5.3, BIC was also estimated manually, using the measurement tools available in CTAn analyser. About 10-15 images were selected from each individual bone scan. The area between the 1<sup>st</sup> and 6<sup>th</sup> implant threads embedded in bone was selected. The BIC was

estimated 10 pixels away from the implant surface, in order to avoid any artifacts, using the following equation:

$$\text{BIC} = \frac{\text{the bone in contact with implant surface}}{\text{the perimeter of the implant}} \times 100$$



**Figure 5.3** A sample of micro-CT reconstructed images used for BIC estimation on [a] human mandibular bone and [b] porcine bone femur.

## 5.3 Results

### 5.3.1 BMD.

Human mandibular bone: The average range of the BMD estimated for the cortical bone was 1.105-1.468 g/cm<sup>3</sup>, while the range for the trabecular bone BMD was 0.43-0.854 g/cm<sup>3</sup> (see Table 5.1).

Porcine bones: The average range of the BMD estimated for cortical bone was 1.736-2.21 g/cm<sup>3</sup>, while the average range of the trabecular BMD was 0.601-0.982 g/cm<sup>3</sup> (see Table 5.1).

**Table 5.1** BMD values estimated from the micro-CT images of the bone samples.

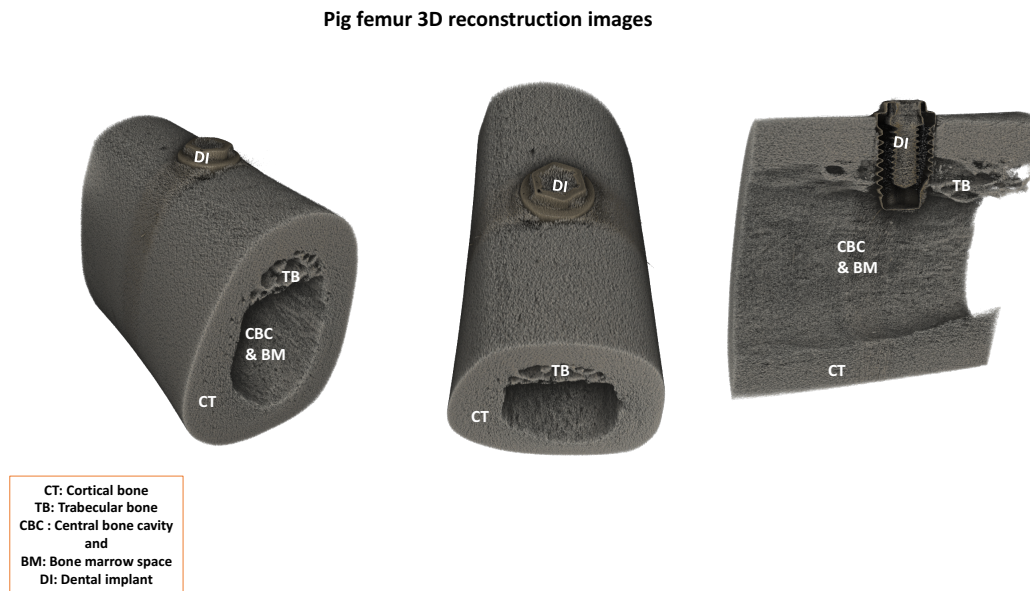
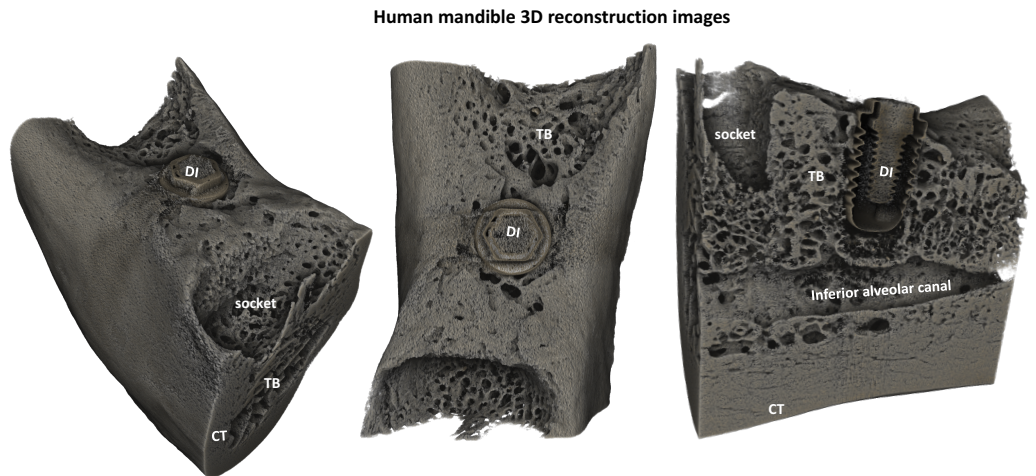
| Sample number | Human bone |            | Pig bone |            |
|---------------|------------|------------|----------|------------|
|               | Cortical   | Trabecular | Cortical | Trabecular |
| 1             | 1.105      | 0.539      | 2.21     | 0.982      |
| 2             | 1.468      | 0.854      | 1.736    | 0.601      |
| 3             | 1.25       | 0.624      | 1.825    | 0.764      |
| 4             | 1.16       | 0.6        | 1.624    | 0.624      |
| 5             | 1.34       | 0.43       | 2.04     | 0.83       |
| Mean          | 1.26       | 0.61       | 1.866    | 0.76       |
| SD            | 0.145      | 0.156      | 0.2      | 0.16       |

### 5.3.2 Morphometric analysis of the bone-to-implant interface.

The reconstructed 3D micro-CT images demonstrated distinct differences in the microarchitecture of the bone samples, as can be observed in Figure 5.4. The porcine femur bone



sample had a thick, dense cortical bone, and a slightly porous, thin trabecular bone layer surrounding a wide medullary cavity. Most of the dental implant was embedded in the dense cortical bone structure. The apical region of the implant seemed to be located in the medullary cavity. The human mandible bone sample had a dense cortical layer (less dense than cortical femur bone) surrounding a porous trabecular structure. Most of the dental implant was surrounded by trabecular bone. The apical part of the implant, however, was in an empty space.

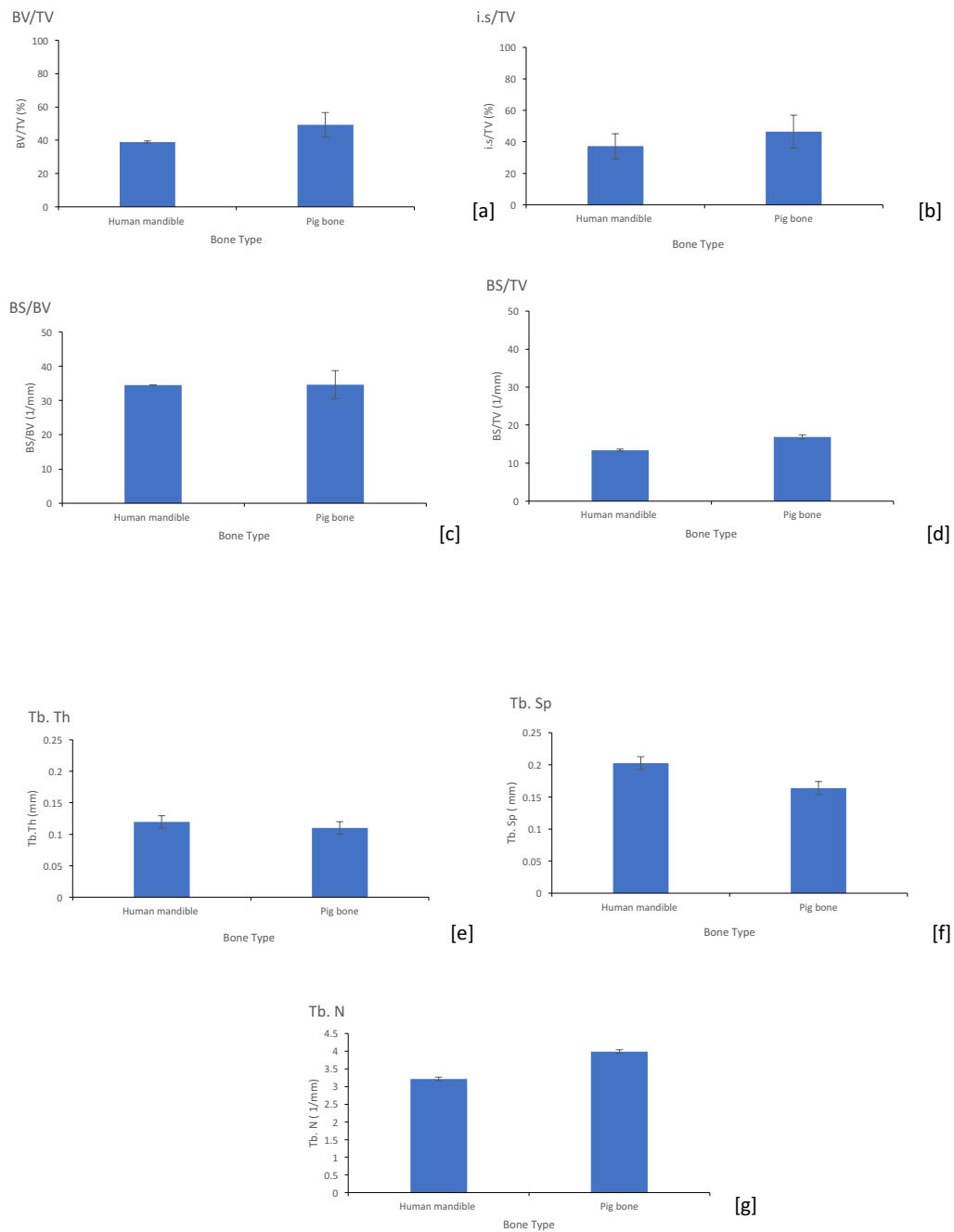


**Figure 5.4** 3D reconstruction of the human and porcine bone samples.

The 3D morphometric analysis results are presented in Table 5.2 and Figure 5.5. The porcine bone had higher BV values compared to the human bone sample. The human bone sample, however, had a significantly higher TS value compared to the porcine bone samples. Both samples had similar TV. The porcine bone samples had significantly higher BS and i.S values compared to the human bone samples (see Table 5.2). The porcine bone samples also had higher bone volume fraction (BV/TV) and bone surface density (BS/TV) values compared to the human mandibular bone samples, but the differences were not statistically significant. Both human and porcine bone samples had similar specific bone surface BS/BV values. For purposes of the trabecular bone analysis, Tb.N was estimated by averaging the number of trabeculae in a 1mm diameter sphere. The porcine bones had a slightly higher number of trabeculae per mm in 3D than did the human bone samples. Trabecular thickness (Tb.Th) and trabecular separation (Tb.Sp) provided mean values to quantify trabecular microarchitecture. Both samples had similar Tb.Th, while the human bone had the highest Tb.Sp value.

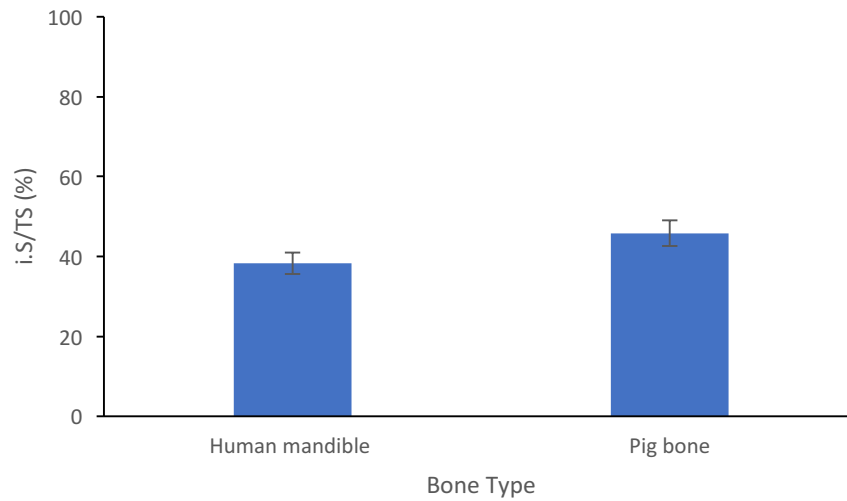
**Table 5.2** Micro-CT 3D analysis parameters.

| $\mu$ CT parameters    | Human mandible |       | Pig bone |        |
|------------------------|----------------|-------|----------|--------|
|                        | Mean           | SD    | Mean     | SD     |
| TV (mm <sup>3</sup> )  | 26.04          | 3.39  | 25.99    | 2.31   |
| BV (mm <sup>3</sup> )  | 10.14          | 1.5   | 12.72    | 0.78   |
| TS (mm <sup>2</sup> )  | 294.8          | 47.46 | 201.56   | 28.65  |
| BS (mm <sup>2</sup> )  | 350.3          | 52.23 | 438.98   | 25.42  |
| i.S (mm <sup>2</sup> ) | 109.7          | 18.68 | 140.79   | 11.652 |



**Figure 5.5** Results of the 3D morphometric analysis of the micro-CT images.

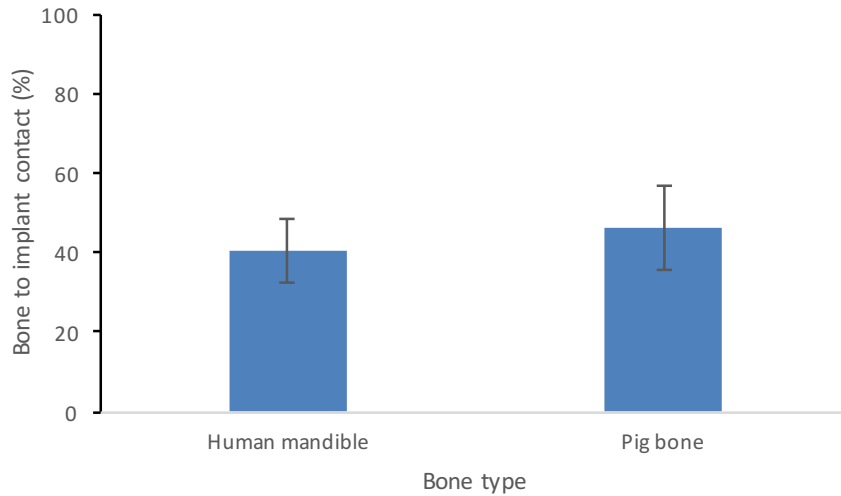
The i.S/TV (BIC) estimated from the 2D micro-CT analysis is illustrated in Figure 5.6. The porcine bone samples had a slightly higher i.S/TV value but the difference relative to the human bone samples was not statistically significant.



**Figure 5.6** i.S/TS ratio estimating BIC obtained from 2D analysis of the micro-CT images.

### 5.3.3 *Measuring BIC manually from micro-CT images.*

The porcine bone BIC values were slightly higher than the human bone values, as can be observed in Figure 5.7. The BIC values obtained from both the 2D (i.S/TV) and 3D (BV/TV) analyses corresponded closely with the manually estimated BIC. The human and porcine bone BIC values themselves for the different measurements were also very similar.



**Figure 5.7** Manual estimation of BIC values obtained from micro-CT images.

## 5.4 Discussion

In this chapter, our objective was to assess the human and porcine bone models, using micro-CT imaging in order to: (1) estimate the BMD and (2) perform a morphometric analysis of the bone-to-implant interface, the results of which are discussed in the narrative below.

#### 5.4.1 *Estimating BMD.*

As mentioned in Chapter 2, the density of bone available for implant placement influences the healing and overall outcome of dental implant treatment (Branemark et al., 1977; Drago, 1992). For this reason, it is important to evaluate bone density prior to implant placement. BMD is defined as the volumetric density of CaHA in a biological tissue in  $\text{g/cm}^3$ . It is calibrated by means of phantoms with known density of CaHA. BMD measured by micro-CT relates to the amount of bone within a mixed bone-soft tissue region, but does not give information about the material density of the bone itself (Bruker-MicroCT, 2005).

The reconstructed grey-scale intensity of each image voxel measured by micro-CT is not directly related to mass density alone, so it might not correspond with measurements of weight per volume ( $\text{g/cm}^3$ ). The BMD unit is measured in terms of x-ray absorption, the attenuation coefficient (AC) in units of  $1/\text{distance}$  ( $\text{mm}^{-1}$ ). This is determined both in terms of mass density and elemental composition of the material (Bruker-MicroCT, 2005). However, when the x-ray absorption of a material involves only one specific material, the measured x-ray AC can be related to the mass density of that material. BMD is calibrated by assuming that the x-ray attenuation within the mineralized material (e.g., bone) is dominated by and can be approximated as the x-ray attenuation of the mineral compound CaHA. Phantoms with known mass concentrations of CaHA are then used. By relating these mass concentrations of CaHA with the measured x-ray AC in the micro-CT image, a calibration connecting the two quantities can be created. Where the density of CaHA in  $\text{g/cm}^3$  can be estimated from the micro-CT measured x-ray AC in a mineralized tissue (Bruker-MicroCT, 2005). Miguel-Sanchez et al. (2015) suggested that BMD should be measured by relating the values obtained from a CT scan to a calibration bone phantom of predetermined

mineral density. Overestimation of BMD may result from lack of precision between the measured BMD and calibrated BMD. This is more worrisome in *in vivo* densitometries (Blake, 1996). Because of the difference in the actual density between rat bone phantoms and the human and porcine bone samples in our study, a 20% range of error was expected in the results.

Although BMD estimated by micro-CT is not directly related to mass density alone, the estimated BMD values for the bone samples corresponded closely well with the apparent density values estimated in Chapter 4 (see Tables 4.1 and 5.1). BMD and apparent density values for the porcine bone samples were higher than those of the human mandibular bone samples. A positive correlation was seen between the natural frequency values and the BMD values for both bone samples, indicating that natural frequency values increase with increases in BMD.

Micro-CT is used in research to assess the BMD of small animals such as mice, rats, and rabbits (Bouxsein et al., 2010). However, limited data are available in the literature for human mandibular bone and porcine bone BMD micro-CT estimation (Lee & Jasiuk, 2014; Mulder et al., 2004). Micro-CT is a fairly new technique that can be difficult to use in practice. The high level of x-ray exposure and the small field of view have made its use in clinical studies limited. Other methods used to measure human bone density include densitometric measurements of panoramic and periapical radiographs, and advanced tools such as Dual Energy X-Ray Absorptiometry (DEXA) (Carey et al., 2007; Drage et al., 2007; Elowsson et al., 1998; Gulsahi et al., 2010), computed tomography (CT) (Turkyilmaz et al., 2008; Turkyilmaz & McGlumphy, 2008a, 2008b; Turkyilmaz et al., 2009; Turkyilmaz et al., 2007a; Turkyilmaz et al., 2006; Turkyilmaz et al., 2007b), and cone-beam computed tomography (CBCT) (Kang et al., 2016; Ko et al., 2017; Naitoh et al., 2010; Pagliani et al., 2013; Soardi et al., 2014; Wang et al., 2013). These measurements,



especially DEXA and CBCT, involve lower levels of exposure to radiation, as well as easier access to the technologies, relative to micro-CT.

A study by Lee and Jasiuk (2014), is one of the few that estimated the BMD for porcine femur head samples using micro-CT. The BMD value was estimated to be  $1.18\text{g/cm}^3$  ( $\text{SD}\pm 0.089$ ) for old trabecular frozen porcine femur samples. The values obtained by Lee and Jasiuk (2014), despite the experimental setup differences, are very close to the values obtained in this study for the trabecular porcine bone femur.

#### 5.4.2 *Morphometric analysis of the bone-to-implant interface.*

The 3D reconstructed micro-CT images demonstrated distinct differences in the microarchitecture of the bone samples, as reflected in Figure 5.4 and described in the results. The dental implant placed in the porcine bone had significantly higher natural frequency values (as mentioned in the preceding chapter) than was the case for the implant placed in the human mandibular bone. In addition to the differences in terms of BMD, the variation in natural frequency can also be explained by the microarchitecture of the bones. In the porcine femur, the dental implant was embedded in the dense cortical bone structure. The dental implant in mandibular bone, however, was placed in the porous trabecular bone structure.

Several studies have attempted to evaluate the morphometric parameters of human jaw bones (Bertl et al., 2015; Blok et al., 2013; de Oliveira et al., 2012; Fanuscu & Chang, 2004; Kim et al., 2013), porcine bones (Mulder et al., 2004; Mulder et al., 2005), and the peri-implant area (Akca et al., 2006; Dias et al., 2015; Fang et al., 2014; Feng et al., 2016; Hsu et al., 2013; Kang et al., 2016; Lin et al., 2009; Neldam & Pinholt, 2014; Park et al., 2005; Schouten et al., 2009).

In the current study, for parameters related to bone volume, porcine bone had slightly higher BV and BV/TV values. For the surface parameters, the porcine bone had a significantly higher BS value compared to the human bone value. Human bone had higher TS values, but the BS/BV values fell in the same range. The BS/TV, however, was slightly higher for the porcine bone, compared to human bone. The internal arrangement of trabeculae is estimated by Tb.Th, Tb.N, and Tb.Sp. Although the Tb.Th was the same for both samples, the Tb.N was higher for the porcine bone sample, while the Tb.Sp was higher for the human mandible sample.

According to de Oliveira et al. (2012) denser bones may have lower Tb.N, Tb.Sp, and Tb.Th values. They can also have well-connected trabeculae, where the trabeculae are in close proximity, forming a compact trabecular bone structure. In addition, denser bones are thought to be related to high bone-volume parameters and low bone-surface parameters, either because the trabeculae are thick or the bone consists of a wide plate-like trabecular pattern, unlike low-density bone. Low-density bones can have high Tb.N, Tb.Sp, and Tb.Th values, but they can have larger trabecular spaces between parallel, plate-like trabeculae, as well as poorly connected bone. This is not consistent with our findings, however. Bone microarchitecture is complicated, as it involves numerous trabecular structure configurations. As each sample (human and bone), as illustrated in Figure 5.4, has unique bone patterns, comparing their microstructures is complicated and difficult, as mandibular bone is very different in structure and architecture from femur bone. In addition, our study samples had implants, which might have resulted in overestimation of the morphometric parameters. Moreover, our samples were derived from old dry bones, which lose a lot of the microarchitecture present in fresh or *in vivo* bones.

Various studies (Fanuscu & Chang, 2004; Moon et al., 2004; Nagele et al., 2004; Rebaudi et al., 2004) have noted that examining bone microarchitecture might help us understand more about its mechanical characteristics and its impact on implant therapy. *In vivo* studies can help determine the actual role of the morphometric parameters on the long-term success of implant therapy (de Oliveira et al., 2012).

Micro-CT provides BIC information, with less sample preparation than histology slides. However, the presence of implant artifacts and low-image quality, compared to histomorphometric analysis, represent major disadvantages in the use of micro-CT (Bouxsein et al., 2010; Dias et al., 2015; Schouten et al., 2009).

Micro-CT artifacts occur around the metal implant because of the high x-ray absorption and especially because of the intense beam hardening caused by metal. These artifacts are usually caused by acute beam hardening, resulting from the high-Z absorption in the metal. The severity of these artifacts depends on the implant substance. In most cases, the implant is made of titanium, a light metal ( $Z=22$ ). The metal attenuation of x-rays is much higher than bone, but is low enough to allow scanning of titanium implants in bone using micro-CT scanners (Bruker-MicroCT, 2015). The scanned implants in our study exhibited a degree of scatter, reduced by filtering the small noise particles and by removing pores in the implant. In addition, the 10-pixel area closest to the implant surface was excluded from the calculation.

The BIC in this study was estimated using 3D morphometric analysis (BV/TV and i.S/TV), 2D analysis (i.S/TV), and manually calculating the BIC of the 2D images. Each of these techniques yielded similar results. In all samples, porcine bone BIC was higher than for human bone, as the dental implant placed in cortical porcine bone has a higher BMD compared to the implant placed

in lower BMD human trabecular bone. In addition, the frequency values estimated for human and porcine bone samples exhibited a positive correlation with the BIC values. Moreover, the BMD values were positively correlated with BIC.

Many studies have compared histomorphometric measurements and micro-CT morphometrics with regard to determining BIC values. Some studies suggest that micro-CT offers a good representation of BIC (Akca et al., 2006; Bernhardt et al., 2012; Fang et al., 2014; Feng et al., 2016; Hsu et al., 2013; Kang et al., 2016). Others suggest that micro-CT is not accurate in estimating BIC, but could complement histomorphometric analysis (Dias et al., 2015; Park et al., 2005; Schouten et al., 2009; Tamminen et al., 2011). Some studies have suggested removal of the implant before scanning to reduce the artifacts (Ribeiro-Rotta et al., 2014). These studies all varied in terms of sample size, type of bone, and scanner set up, making it difficult to compare them.

A study by Fu et al. (2017) evaluated whether ISQ could be used to predict bone quality. The authors concluded that ISQ values had a weak correlation with bone type when assessed by micro-CT in the maxilla. They also cautioned regarding the use of the Osstell ISQ<sup>®</sup> as a tool to evaluate bone quality at the implant site, especially in the mandible. Other studies, by Akca et al. (2006) and Dias et al. (2015), concluded that insertion torque value (ITV) was more sensitive than ISQ in terms of revealing biomechanical properties at the bone–implant interface.

In summary, micro-CT offers a non-destructive method for analyzing the microarchitecture of bone and estimating BMD. Additional studies are needed, however, to validate their use for peri-implant analysis. Increasing our knowledge about the microarchitecture of peri-implant bones may prove helpful in evaluating the reliability of implant stability measurements.

## 5.5 Conclusions

Considering the limitations inherent in this study, as well as the limitations uncovered by this study, in this chapter we conclude the following:

- Porcine bones had higher estimated BMD values compared to human BMD values.
- The estimated BMD from micro-CT fell within the range of the apparent density values.
- The morphometric analyses of the human and porcine bone structures in this study are difficult to compare. As each sample has unique bone patterns, comparing their microstructures is complicated and challenging, as mandibular bone is very different in structure and architecture from femur bone. In addition, the fact that the samples we worked with had implants might have resulted in overestimating the morphometric parameters.
- The BIC estimated by the three different methods—BV/TV, i.S/TV, and manual assessment—yielded similar results.
- The BIC for porcine bone was slightly higher than the value for human bone.
- The BMD was positively correlated to BIC and the natural frequency values, for both types of bone.

The following chapter will discuss how boundary conditions and experimental setup can significantly affect implant stability measurements.

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## **CHAPTER 6. EFFECTS OF EXPERIMENTAL SETUP AND BOUNDARY CONDITIONS ON DENTAL IMPLANT STABILITY TESTING *IN VITRO***

### **6.1 Introduction**

In the preceding chapters, to observe how effectively Sawbones<sup>®</sup> and porcine bones could be used to accurately represent human mandibular bones, we compared the different types of bone models (i.e., porcine bone, human mandibular bone, and Sawbones<sup>®</sup>), using the Osstell ISQ<sup>®</sup>, and the Periotest<sup>®</sup> devices, and experimental modal analysis (EMA), in addition to measuring apparent bone density. We also used micro-computed tomography (micro-CT) to estimate bone mineral density (BMD) and morphometric parameters of both porcine and human bone samples, concluding that porcine bone models probably tend to result in overestimating implant stability. The Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV were unable to differentiate between implants in different types of bone models, suggesting their unreliability for this purpose. Although Sawbones<sup>®</sup> may overall offer a good model for representing human bone, measurements resulting therefrom should

be considered with caution, as Sawbones<sup>®</sup> do not accurately reflect actual bone biomechanics. In addition, the estimated BMD (for human and porcine bone) from micro-CT fell within the range of the apparent density values. The morphometric analyses of the human and porcine bone structures were difficult to compare. As each sample has unique bone patterns, comparing their microstructures is complicated and challenging, as mandibular bone is very different in structure and architecture from femur bone. In addition, the fact that the samples we worked with had implants might have resulted in overestimating the morphometric parameters. Moreover, the BIC estimated by the three different methods—BV/TV, i.S/TV, and manual assessment—yielded similar results and was positively correlated to BMD and the natural frequency values, for both types of bone.

As discussed in section 2.6 of Chapter 2, many studies have been carried out to assess implant stability, for the purpose of arriving at criteria for clinical use (Bilhan et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016; Sennerby et al., 2015; Tabassum et al., 2010; Wang et al., 2015; Yamaguchi et al., 2015). The use of *in vitro* test models in these studies has often involved the placement of dental implants in Sawbones<sup>®</sup> (Ahmad & Kelly, 2013; Ahn et al., 2012; Giovanni et al., 2011; Hsu et al., 2013; Tabassum et al., 2010; Wang et al., 2015). ). In addition, we mentioned that to better control the results yielded from the use of test models, special attention must be paid to two particular issues: boundary conditions and experimental setup.

Despite the importance of conditions relating to boundary and experimental setup, the literature includes very few studies addressing these two important aspects. Potential issues

relating to experimental setup and boundary conditions for *in vitro* tests have for the most part been ignored, as little discussion has taken place in the dental literature on the effects of boundary conditions and experimental setup. Nor is there any standard protocol as to how the conditions relating to boundary and experimental setup should be managed for purposes of *in vitro* dental implant stability testing. One of the few groups of studies to address boundary conditions is that of Huang et al., who conducted a series of *in vitro* (2003a; 2002; 2000a; 2000b) and *in vivo* (2003b) experiments using EMA. The studies by Huang et al. (2003a; 2003b; 2002; 2000a; 2000b), however, are qualitative as discussed in Chapter 2.

For development of an implant stability clinical assessment tool, it is critical that the implant test model response be well calibrated, both theoretically and experimentally. For example, natural frequency values of the test model should be measured experimentally and simulated numerically. Moreover, the measured and simulated results should agree quantitatively. Only then can the test model serve as an accurate benchmark for evaluating existing and novel assessment tools for implant stability. To achieve this goal, in Chapter 3, we used EMA to measure natural frequency values of a test model, with a dental implant placed in a Sawbones<sup>®</sup> block. The same test model was also simulated numerically via FEA. The EMA measurements and the FEA predictions correlated well quantitatively. Therefore, the test model was used to calibrate other assessment tools, such as the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> devices. The effects of boundary conditions and experimental setup on the test model call for investigation, in order to define, delineate, and measure in great detail to confirm accuracy and provide confidence in the test model.

The need for a more reliable method for assessing implant stability, in the context of the lack of published studies in this area, prompts us in this chapter to demonstrate how experimental

setup and boundary conditions affect implant stability measurements. Our study of experimental setup involves the use of dental implants of various lengths, as well as the use of two different types of abutments. Our study of boundary conditions focuses on side (lateral) anchorage in a vise (as explored in Chapter 3), as opposed to a bottom-positioned anchorage of the Sawbones<sup>®</sup> blocks using adhesive tape. To calibrate the test model, natural frequency values were measured by means of EMA and confirmed numerically through use of FEA, for various combinations of both boundary conditions and experimental setup. For purposes of benchmarking through use of the test model, dental implant stability was estimated using two commercial devices: the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>. Finally, a parametric study was performed through the use of FEA, to identify the cause of considerable variations observed in the measured natural frequency values.

## **6.2 Materials and Methods**

### *6.2.1 Simulated jawbone.*

Three different Sawbones<sup>®</sup> densities were used: hybrid blocks (40/20-PCF), high-density (40-PCF) blocks, and low-density blocks (15-PCF) as described in Chapter 3, section 3.2.3.

### *6.2.2 Dental implants.*

Branemark Mk III implants 4 mm in diameter and lengths of 7, 8.5, 10, 11.5, 13, 15, and 18 mm were placed in the center of each block following manufacturer's surgical protocol seating to 45 N·cm as described in Chapter 3, section 3.2.4.

For the Periotest<sup>®</sup> and experimental modal analysis (EMA) tests, a locator abutment (LA) 7 mm in length and an impression coping (IMP) abutment 12 mm were used. A 10 N·cm torque was used to tighten the abutment with Branemark System Torque Control.

### 6.2.3 *Boundary conditions.*

Sawbones<sup>®</sup> samples were secured via two types of boundary conditions: side-anchorage (as used in Chapter 3) and bottom-positioned anchorage. For the side-anchorage, the lower 13-mm part of each Sawbones<sup>®</sup> block was firmly clamped by a vise that was bolted to an isolation table to prevent possible external disturbances. Therefore, two parallel sides of the Sawbones<sup>®</sup> block were in a fixed boundary condition. For the bottom-positioned anchorage, the bottom surfaces of the Sawbones<sup>®</sup> samples were attached to the top surface of a steel block (63.5 X 63.5 X 14.5 mm in dimension) using carpet tape. Specifically, the Sawbones<sup>®</sup> samples were pressed onto a central taped portion of the steel block with hand pressure. In addition, we made sure that the carpet tape was placed properly with no air bubbles and the Sawbones<sup>®</sup> was fixed on to the steel block with no apparent gaps or rocking. Finally, the steel block was clamped firmly in the vise. In this setup, the bottom surface of the Sawbones<sup>®</sup> samples was in a fixed boundary condition.

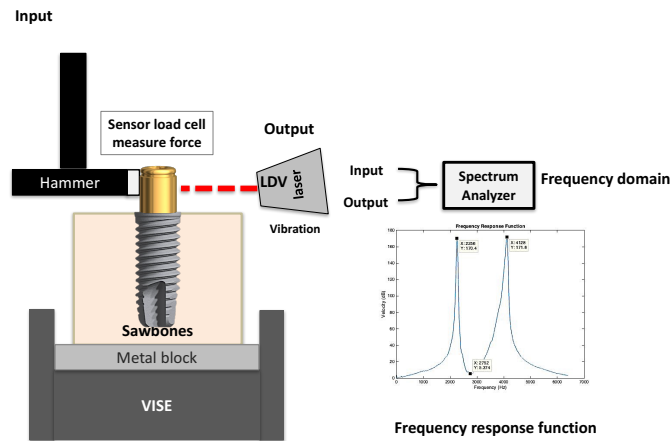
### 6.2.4 *Measurement methods.*

Implant stability of each test block was measured through use of the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and experimental modal analysis (EMA). For the Osstell ISQ<sup>®</sup>, SmartPeg<sup>®</sup> (type I) was screwed on the implant at the fixture level with finger pressure and measurements were taken mesio-distally (MD).

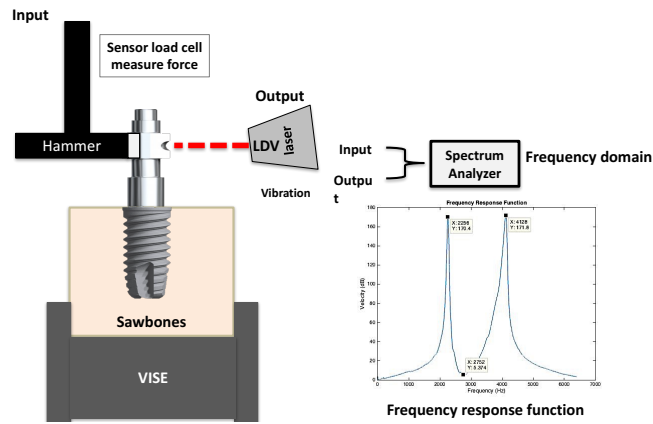
For the Periotest<sup>®</sup>, the samples were tested with the device tip placed perpendicular to the access of the implant and a few millimeters away. The average of three readings was taken for each set of measurements and the abutment was removed and re-attached between measurements.

For EMA, the setup is described in detail in Chapter 3; see Figure 6.1. Based on the measured force and velocity, the spectrum analyzer calculated frequency response functions, from which natural frequency values of the implant-Sawbones<sup>®</sup> test models were extracted.





[a]



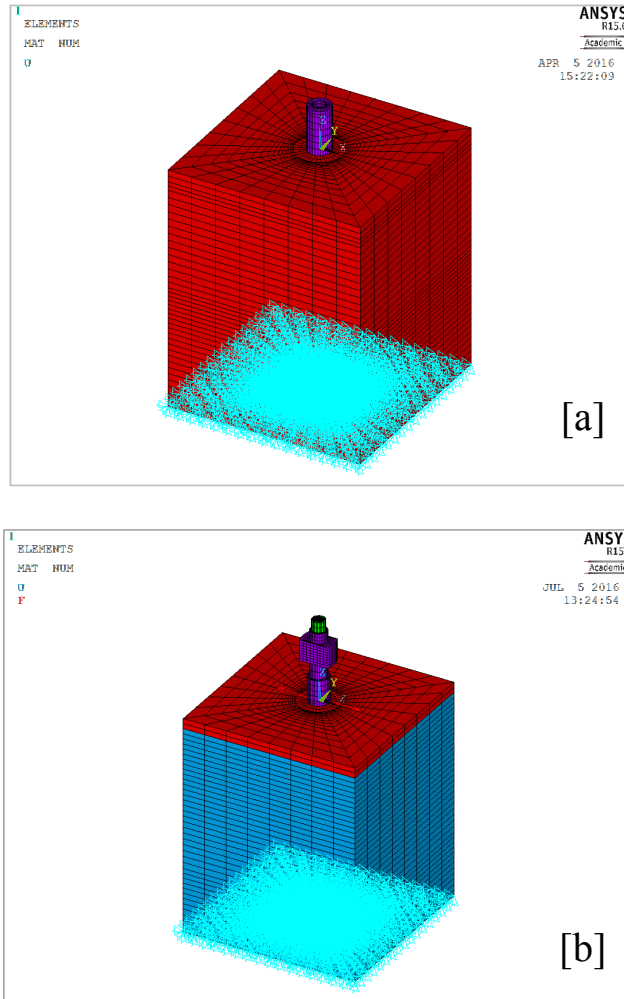
[b]

**Figure 6.1** Experimental modal analysis (EMA) setup involves a hammer (for measuring force input) and a laser Doppler vibrometer (to measure velocity output). The measured force and velocity are processed by means of a spectrum analyzer, for purposes of obtaining frequency response functions. [a] A Sawbones<sup>®</sup> model with a locator abutment (LA) under bottom-positioned anchorage boundary conditions. [b] A Sawbones<sup>®</sup> model with an impression coping (IMP) abutment with side-anchorage boundary conditions (copied from Chapter 3 for comparison with [a]).

### 6.2.5 Finite element analysis (FEA).

A 3D finite element model (FEM) was created using ANSYS R-15 (Canonsburg, PA) to simulate the experimental setup and the test results. Figure 6.2 [a] shows an FEA model of an

experimental setup with a bottom-positioned anchorage and an LA. The model was built using SOLID186 elements, which are higher-order, 3D, 20-node solid elements that assume quadratic displacement fields. The FEM consists of three parts: a Sawbones<sup>®</sup> block, a cylindrical implant, and an LA. For the Sawbones<sup>®</sup> block, material properties (e.g., density and Young's modulus) provided by the manufacturer were used to model the tested Sawbones<sup>®</sup> blocks of three different densities (see Table 3.1). The Sawbones<sup>®</sup> block was assumed to be isotropic and had the same size as the test samples. Moreover, the block was fixed at the bottom to reflect the bottom-positioned anchorage boundary conditions imposed in the experiments. To model the implant, one-piece titanium cylinders 7, 8.5, 10, 11.5, 13, 15, and 18 mm in length and 4 mm in diameter were used. Moreover, the cylinders were located at the center of each Sawbones<sup>®</sup> block. The nodes of the cylindrical implants and the Sawbones<sup>®</sup> block were merged at the implant-Sawbones<sup>®</sup> interface to model a no-slip condition (i.e., perfect bonding) between the implant and the Sawbones<sup>®</sup> block. The LA was modeled in the exact dimensions. The material of the LA was titanium alloy. It was connected to the implant via a no-slip condition. After the models were created, a modal analysis was conducted to predict natural frequency values and mode shapes of the simulated test samples. Similarly, an FEM was created to simulate the experimental setup with bottom-positioned anchorage and an IMP abutment; see Figure 6.2 [b].

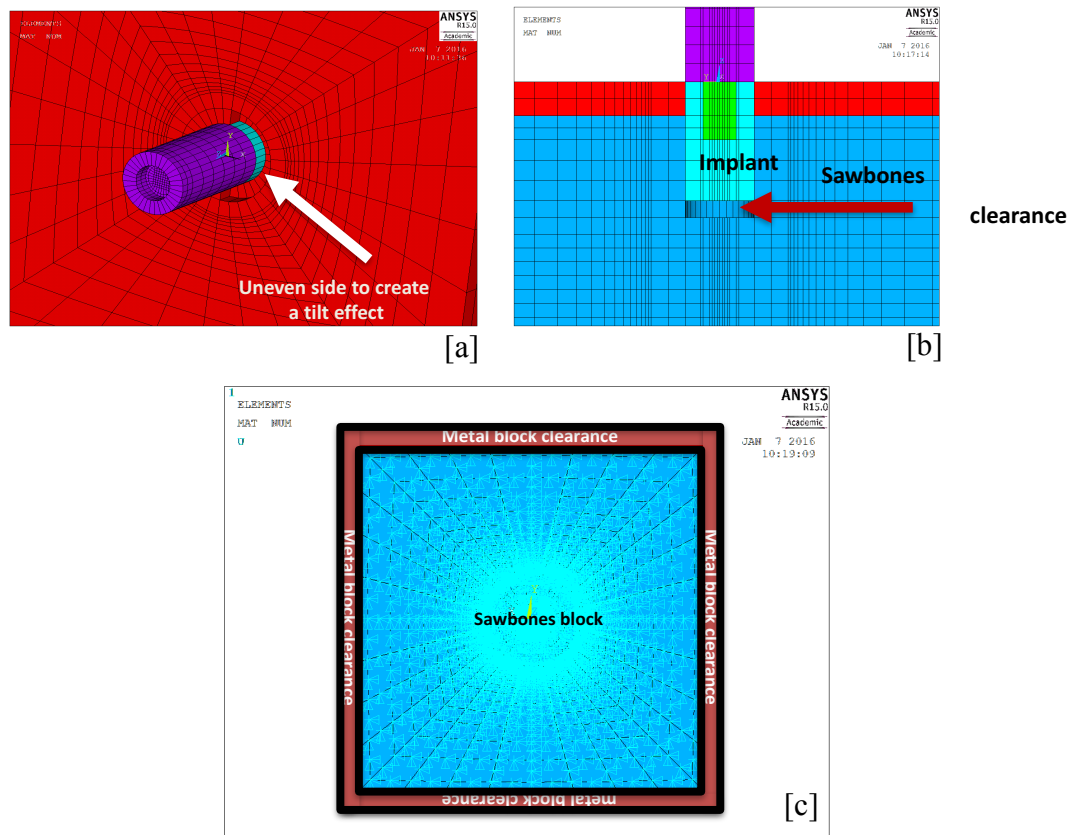


**Figure 6.2** Finite element models under bottom-positioned anchorage boundary conditions: [a] a model with the locator abutment (LA), and [b] a model with the impression coping (IMP) abutment.

#### 6.2.6 Parametric study.

A parametric study was performed through the use of FEA to identify the cause of considerable variations observed in the measured natural frequency values. Basically, several finite

element models were created with slightly different parameters. Five parameters were investigated: (a) tilt of the implant inside the Sawbones<sup>®</sup>, (b) end clearance between implant and Sawbones<sup>®</sup>, (c) contact area of the bottom-positioned anchorage, (d) Sawbones<sup>®</sup> block dimensions, and (e) Sawbones<sup>®</sup> block density. The finite element models associated with the first three parameters are shown in Figure 6.3. In the parametric study, we changed one parameter at a time to investigate the corresponding change in the natural frequency values. For example, we changed the Sawbones<sup>®</sup> block density by 10% to find out the corresponding change of natural frequency values. We then changed the contact area by 10% to find out the corresponding change of natural frequency values. By analyzing results from these models, we identified which parameters affected the natural frequency values most critically and therefore needed to be controlled tightly during experiments.

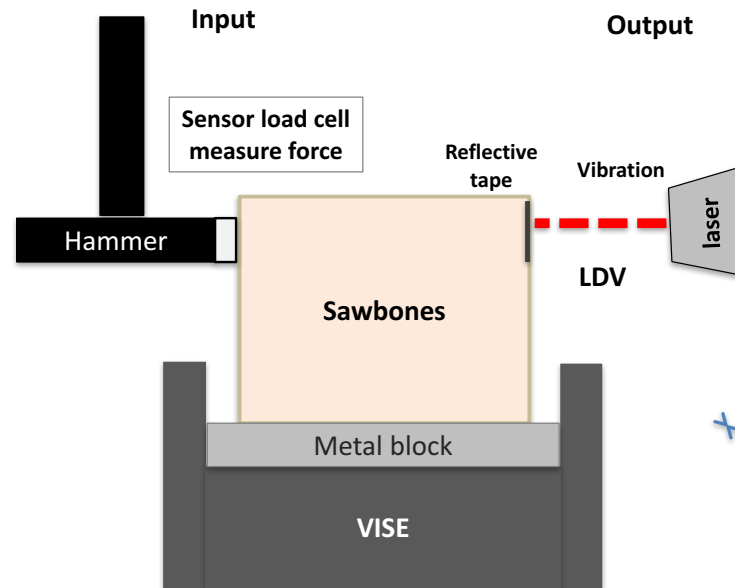


**Figure 6.3** Three finite element models used in the parametric study: [a] uneven sides were created to resemble the effect of the tilt of the abutment resulting from tilt in the dental implant, [b] bottom clearance between implant and Sawbones<sup>®</sup>, and [c] varying the contact area between the Sawbones<sup>®</sup> block and the metal base.

### 6.2.7 Repeatability tests.

According to its specification, Sawbones<sup>®</sup> could have up to  $\pm 10\%$  variations in physical properties. Therefore, a repeatability test was conducted to determine if the  $\pm 10\%$  variations had a measurable effect on the natural frequency values. To eliminate the effect of the dental implants and the bone-to-implant interface, we used intact Sawbones<sup>®</sup> blocks with no drilled-in implants.

Sawbones<sup>®</sup> of the different densities (15-, 40- and 40/20-PCF) taken from various batches were tested; see Figure 6.4. The natural frequency values of these three Sawbones<sup>®</sup> blocks were then measured via EMA and compared for repeatability.



**Figure 6.4** Experimental modal analysis (EMA) setup for the repeatability test. The natural frequency values for Sawbones<sup>®</sup> blocks were measured. Reflective tape was applied to help in reflecting the laser beam.

#### 6.2.8 Statistics.

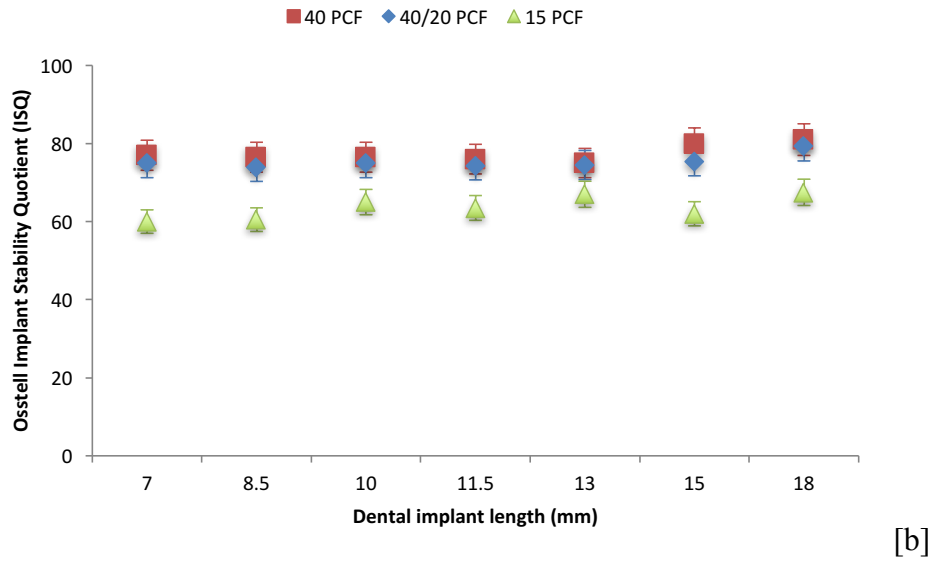
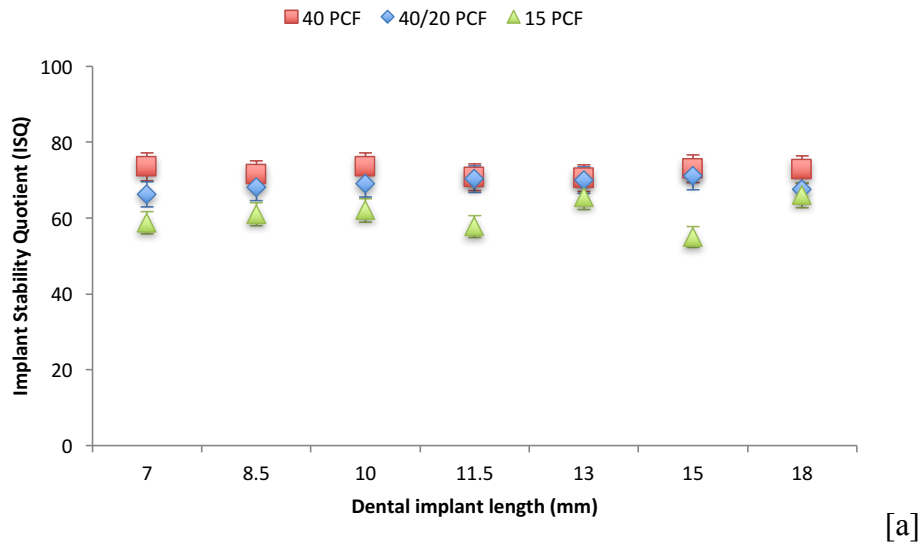
The mean ISQ, PTV, and natural frequency values were calculated as well as their standard deviations (SD). The correlation coefficient between implant length and these values (ISQ, PTV, and natural frequency values) was estimated.

## 6.3 Results

### 6.3.1 *Osstell ISQ*<sup>®</sup>.

The measured ISQ readings represent implant stability according to the manufacturer. Figure 6.5 [a] shows measured ISQ readings for test models with bottom-positioned anchorage and 7 different implant lengths. As shown in Figure 6.5 [a], the ranges for mean mesial-distal (MD) readings were as follows: 66.25-71 for the hybrid blocks, 70.5-73 for the high-density blocks, and 55-66 for the low-density blocks. The ISQ values for the hybrid and high-density blocks fell within the same range while the values for the low-density blocks were lower and more scattered.

To evaluate the effect of boundary conditions, the measured ISQ values above were compared with the ISQ values for test models with side-anchorage boundary condition in Chapter 3 (see Figure 6.5 [b]). In general, the test models with side-anchorage had higher ISQ values. In other words, the side-anchorage made the dental implants seem more stable. The boundary conditions, however, did not significantly affect the trends. For example, models with high-density and hybrid Sawbones<sup>®</sup> blocks had similar ISQ readings, while the models with low-density Sawbones<sup>®</sup> blocks had scattered ISQ readings.



**Figure 6.5** [a] Effects of bottom-positioned anchorage boundary conditions with regard to the relationship between ISQ and implant length. [b] Effects of side-anchorage boundary conditions with regard to the relationship between ISQ and implant length (copied from Chapter 3, for comparison with [a]).



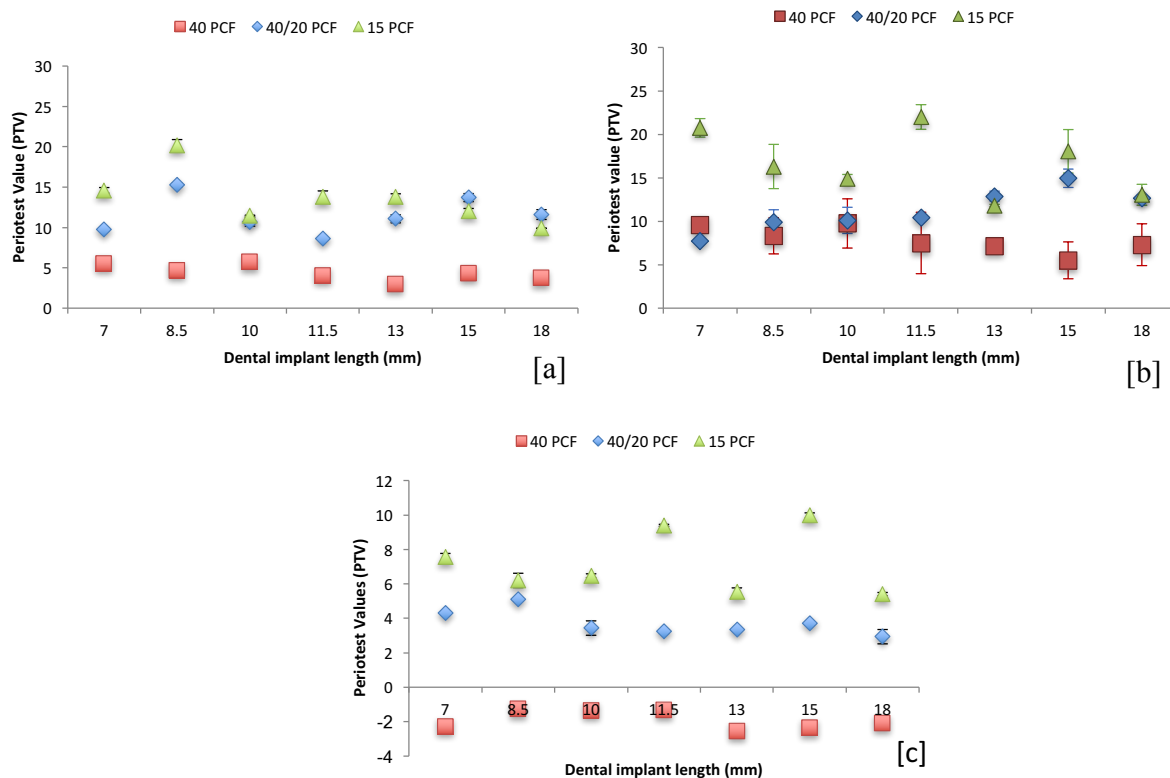
### 6.3.2 *Periotest*<sup>®</sup> PTV.

The measured results for PTV reveal two types of effects: boundary conditions and experimental setup. To study the effects of boundary conditions, we analyzed results from test models using an IMP abutment with bottom-positioned anchorage and side-anchorage. The ranges for mean PTV for the IMP abutment with bottom-positioned anchorage were as follows: 8.6-15.3 for the hybrid blocks, 2.9-5.73 for the high-density blocks, and 9.9-20.2 for the low-density blocks, see Figure 6.6 [a]. The measurements were hard to obtain and were inconsistent. The readings were very scattered, especially for the hybrid and low-density blocks. In general, the high-density blocks had lower readings indicating more implant stability. Moreover, the hybrid and low-density blocks had readings that fell within the same range.

To study the effect of boundary conditions, the above PTV results were compared with PTV from Chapter 3 of test models with IMP abutment and side-anchorage boundary condition. With side-anchorage, the mean PTV for the IMP abutment ranged from 8.6 to 15.3 for the hybrid blocks, 2.9 to 5.73 for the high-density blocks, and 9.9 to 20.2 for the low-density blocks (see Figure 6.6 [b]). Again, only the high-density blocks provided meaningful data. Comparison of Figures 6.6 [a] and [b] shows that test models with bottom-positioned anchorage have lower PTV values implying that they are more stable. This result is contradictory to the observation from ISQ measurements in Figure 6.5, which showed that test models with side-anchorage were more stable.

To study the effect of experimental setup, PTV measurements for test models with LA and bottom-positioned anchorage are shown in Figure 6.6 [c] for comparison. With bottom-positioned anchorage, the mean PTV for the LA ranged between (2.9 and 5.1) for the hybrid blocks, (-2.5 and -1.4) for the high-density blocks and (20.5 and 11.5) for the low-density blocks. For the low-

density blocks, the PTV readings were too scattered to be meaningful. For the high-density and hybrid blocks, the PTV readings were more consistent and showed no correlation to dental implant length. Comparison of Figures 6.6 [a] and [c] also shows that different abutments significantly affected PTV measurements. Test blocks with LA showed more stability than those with IMP abutment at least for the test models with hybrid or high-density blocks, although the stability at the implant-Sawbones® interface remained unchanged.

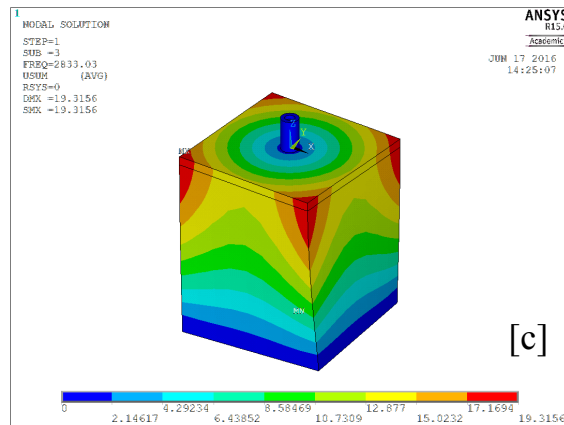
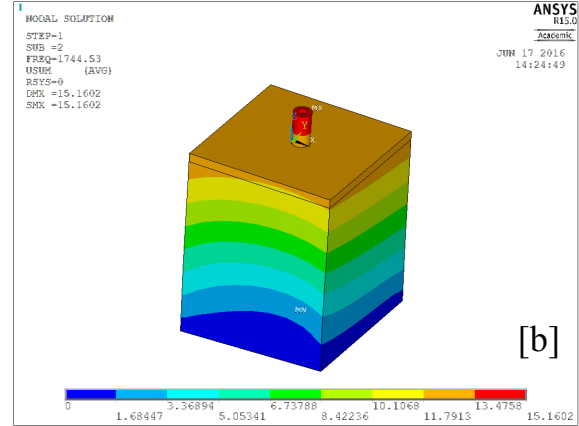
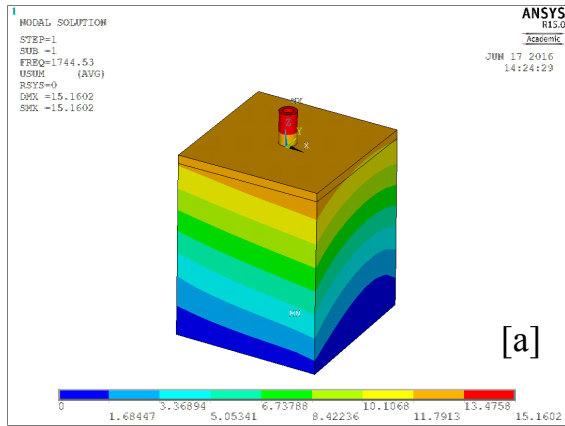


**Figure 6.6** Effects of boundary conditions and experimental setup using PTV measurements: [a] results for test models with IMP abutment (bottom-positioned anchorage) [b] results for test models with IMP abutment (side-anchorage) (copied from Chapter 3, for comparison with [a]), and [c] results for test models with LA (bottom-positioned anchorage).

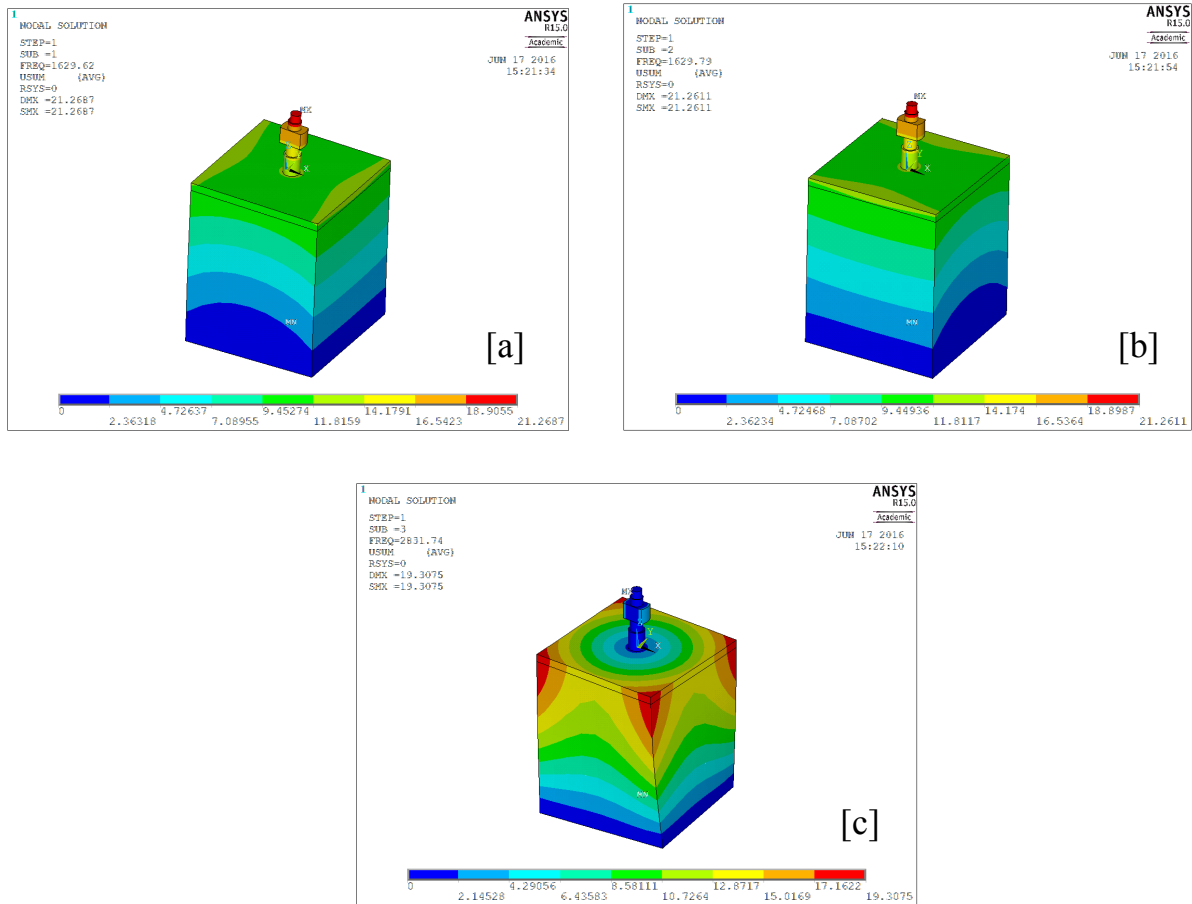
### 6.3.3 *Finite element analysis (FEA).*

The first three major modes with ascending natural frequency values were calculated in FEA for the Sawbones<sup>®</sup> models with LA and IMP abutments and bottom-positioned anchorage shown in Figures 6.2 [a] and [b]. For the model with LA, the first mode represents a forward-and-backward movement of the abutment-implant assembly as demonstrated in Figure 6.7 [a]. The second mode stands for a sideways movement of the abutment-implant assembly (see Figure 6.7 [b]). Since the LA is cylindrical and thus axisymmetric, the natural frequency is the same for the first and second modes. The third mode represents a twisting motion of the Sawbones<sup>®</sup> block as illustrated in Figure 6.7 [c].

For the model with IMP abutment, the results differed because the IMP abutment is not axisymmetric. The first side of the IMP abutment has a larger surface area. The second side is orthogonal to the first side, and has a smaller surface area (see Figure 6.2 [b]). The first mode represents a forward-and-backward movement of the abutment-implant assembly along the normal direction of the first side as demonstrated in Figure 6.8 [a]. The second mode stands for a sideways movement of the abutment-implant assembly along the normal direction of the second side as illustrated in Figure 6.8 [b]. Since the IMP abutment is not axisymmetric, the natural frequency differed for the first and second modes. The third mode represents a twisting motion of the Sawbones<sup>®</sup> (see Figure 6.8 [c]) similar to the case of the LA (see Figure 6.7 [c]).



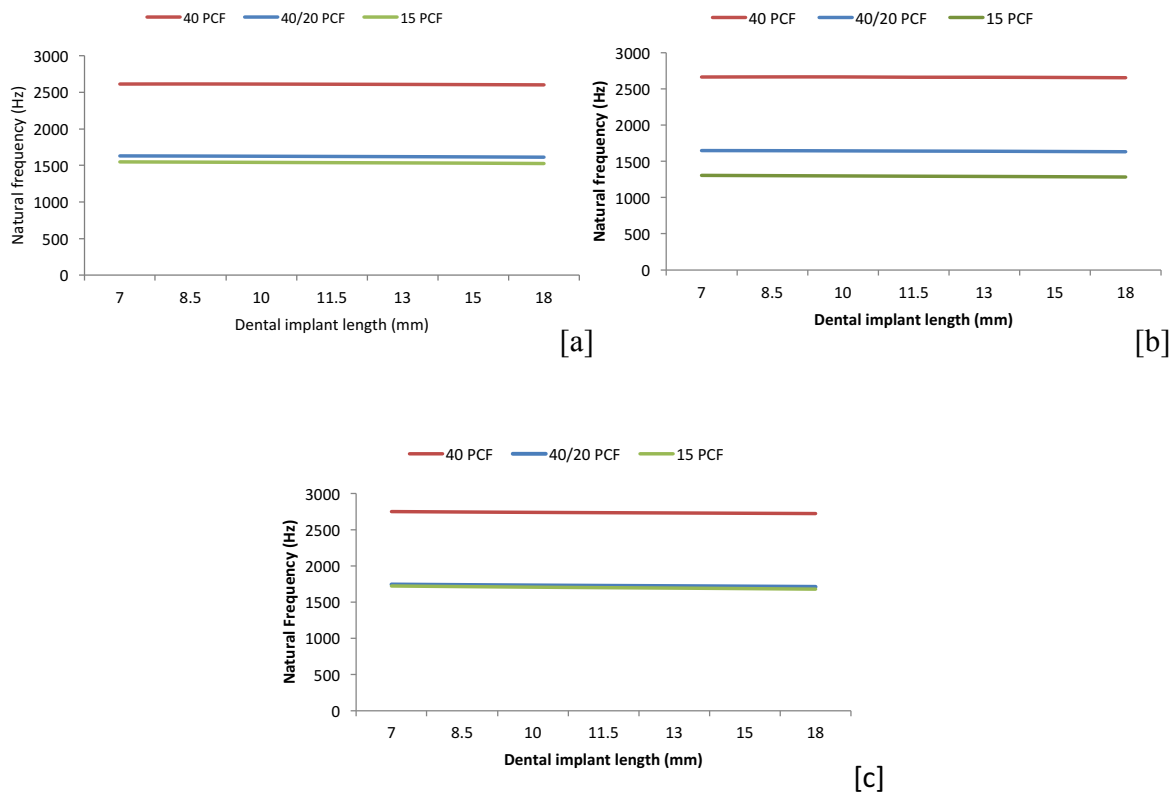
**Figure 6.7** FEA results showing three mode shapes of the test blocks with locator abutment (LA) and bottom-positioned boundary condition: [a] a forward-and-backward movement of the implant-abutment system, [b] a sideways movement of the implant-abutment system (as the LA is cylindrical and thus axisymmetric, the natural frequency is the same for the first and second modes), and [c] a twisting of the Sawbones<sup>®</sup> model.



**Figure 6.8** FEA results showing three mode shapes of the test blocks with impression coping (IMP) abutment and bottom-positioned boundary condition: [a] a forward-and-backward movement of the implant-abutment system, [b] a sideways movement of the implant-abutment system, and [c] a twisting of the Sawbones<sup>®</sup> model.

Figure 6.9 displays FEA predictions of the first natural frequency for [a] models with IMP abutment and bottom-positioned anchorage, [b] is for models with IMP abutment and side anchorage, and [c] models with LA and bottom-positioned anchorage. In general, the change in dental implant length had no effect on the natural frequency values, but the change in the density had a significant effect on the natural frequency values. For the effect of boundary conditions,

Figure 6.9 [a] is compared with the predictions of the IMP abutment with side-anchorage in Figure 6.9 [b]. The test models with IMP abutment had similar frequency values except for the low-density blocks which were significantly higher with bottom-positioned anchorage. For the effect of experimental setup, comparison of Figures 6.9 [a] and [c] indicates that test models with LA had higher natural frequency values than those with IMP abutment.



**Figure 6.9** FEA predictions of the first natural frequency: [a] models with impression coping (IMP) abutment and bottom-positioned anchorage boundary condition, [b] models with IMP and side-anchorage boundary condition (copied from Chapter 3, for comparison with [a]), and [c] models with locator abutment (LA) and bottom-positioned anchorage boundary condition.

#### 6.3.4 *EMA*

Test results of EMA had basically the same trend as the FEA predictions. Figure 6.10 [a] shows the results for the test models with IMP abutment and bottom-positioned anchorage. The measured first natural frequency ranged between (2520 and 2680) Hz for the high-density blocks, (1824 and 1888) Hz for the hybrid blocks, and between (1616 and 1720) Hz for the low-density blocks. Most importantly, the measured natural frequency values did not depend on dental implant length. Moreover, the measured natural frequency values for high-density blocks were significantly higher than those of the hybrid and low-density blocks. These qualitative experimental results agree well with the predictions from FEA. The results from EMA also quantitatively match with the FEA predictions. For the test models with IMP abutment and bottom-positioned anchorage, the FEA predictions of the first natural frequency were (2601- 2612) Hz, (1612-1629) Hz, and (1525-1547) Hz for the high-density, hybrid, and low-density blocks, respectively (see Figure 6.10 [a]). For the high-density, the difference between the EMA measurements and FEA prediction was less than 4%. For the low-density blocks, the difference was around 10%. For the hybrid blocks, however, the difference between the EMA measurements and FEA predictions was significantly higher; roughly around 15%. The cause of the large difference is explained later under “Repeatability test.”

To study the effect of boundary conditions, the measured natural frequency values of the test models with IMP abutment and bottom-positioned anchorage were compared to the test models with IMP abutment and side-anchorage presented in Chapter 3 (see Figure 6.10 [b]). Comparison of the two Figures 6.10[a] and [b] shows that test models with bottom-positioned anchorage had higher natural frequency values than those with side-anchorage. These

experimental results corresponded closely with the FEA predictions shown in Figure 6.10 [a] (for side-anchorage) and 6.10 [b] (for bottom-positioned anchorage), at least qualitatively.

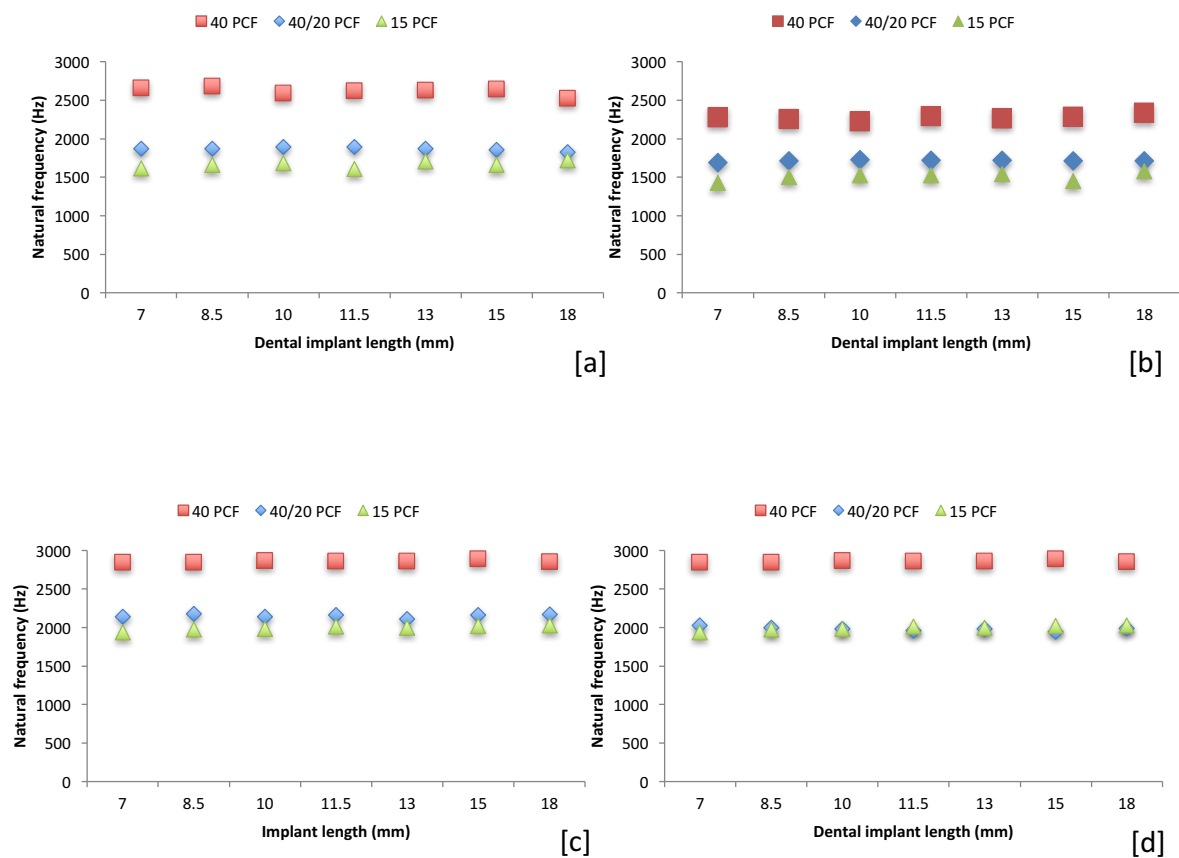
The quantitative agreement between the EMA measurements and FEA predictions, however, was not as high. For the test models with IMP abutment and side-anchorage, the difference between the EMA measurements and FEA predictions was 20%, 5%, and 18% for the high-density, hybrid, and low-density blocks, respectively. The larger difference casts doubt on whether or not the side-anchorage is a good boundary condition for the test models.

To study the effect of experimental setup, measured natural frequencies for the test models with LA and bottom-positioned anchorage are shown in Figure 6.10 [c] for comparison. The measured first natural frequency fell in the range between (2848 and 2888) Hz for the high-density blocks, (2112 and 2176) Hz for the hybrid blocks, and (1936 and 2036) Hz for the low-density blocks. Again, the measured natural frequency values did not have a significant correlation to dental implant length. Comparison of Figures 6.10 [a] and [c] shows that test models with LA had higher natural frequency values than those with IMP abutment. These experimental results agree with the FEA predictions shown in Figure 6.9.

The quantitative agreement between the EMA measurements and the FEA predictions is perplexing. For the test models with LA abutment and bottom-positioned anchorage, the FEA predictions of the first natural frequency were (2722.7- 2750) Hz, (1712-1744.6) Hz, and (1679.3- 1723) Hz for the high-density, hybrid, and low-density blocks, respectively (see Figure 6.9 [c]). The difference between the EMA measurements and the FEA predictions was 5.5%, 14%, and 17% for the high-density, hybrid, and low-density blocks, respectively.



The comparison of EMA measurements and FEA predictions led to several important questions. First, which boundary condition is better? Second, what caused the large differences in the natural frequency values? Third, what can be controlled better in experiments to provide a good proving ground? These questions motivated the following parametric study and repeatability tests.



**Figure 6.10** EMA measurements of the first natural frequency: [a] test models with IMP abutment and bottom-positioned anchorage, [b] test models with IMP abutment and side-anchorage (copied from Chapter 3, to compare with [a]), [c] and [d] test models with LA and bottom-positioned anchorage in which the hybrid (40/20-PCF) blocks were obtained from two separate Sawbones® batches.

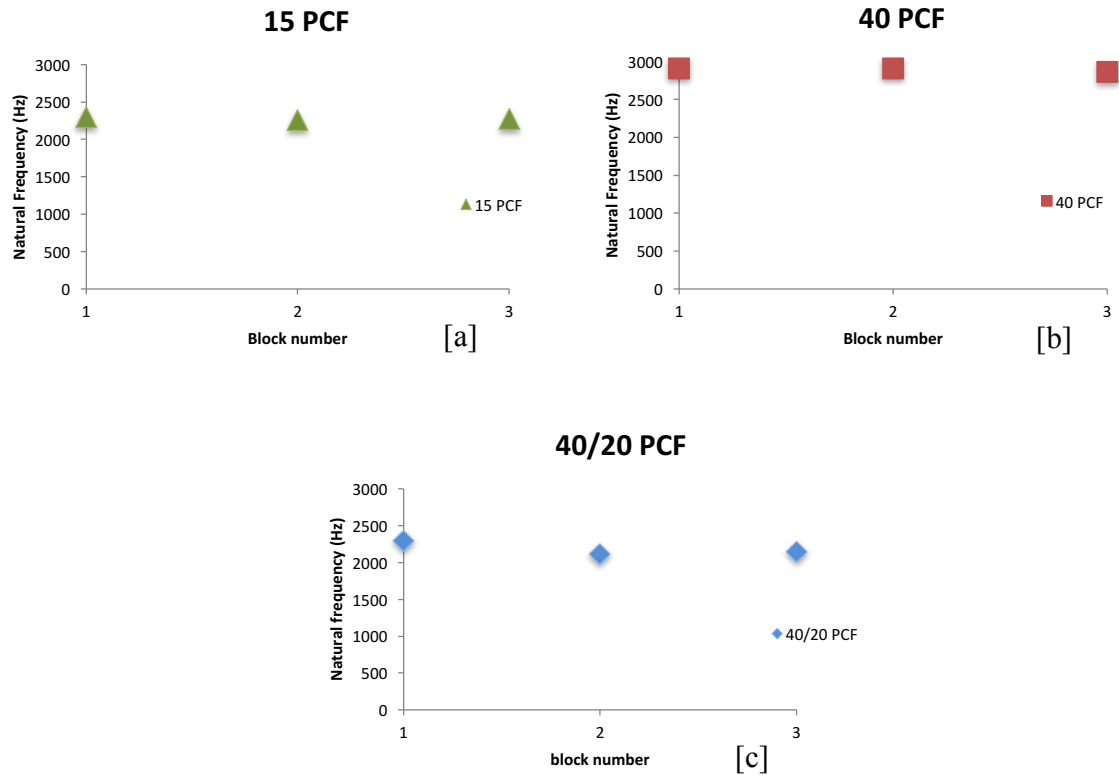
#### 6.3.5 *Parametric study.*

One powerful capability of FEA is the identification, through accurate parametric analysis, which parameters are sensitive and need to be carefully controlled in experiments. The reference point used in the parametric study was the high-density Sawbones<sup>®</sup> blocks. Varying the tilt angle by 1° caused the first natural frequency to shift by 1 Hz. Varying the end clearance between the implant and Sawbones<sup>®</sup> by 1 mm caused the first natural frequency to shift by 1 Hz. Varying the contact area of the bottom-positioned anchorage (i.e., area between the Sawbones<sup>®</sup> and the metal base) by 10% led to a 100-Hz change in the first natural frequency. Varying the block dimensions by 10% caused the first natural frequency to shift by 1 Hz. Finally, varying the density of the Sawbones<sup>®</sup> blocks by 10% caused the first natural frequency to shift by 100 Hz.

From the parametric study, we concluded that the contact area of the bottom-positioned anchorage was the most critical parameter that should be highly controlled in setting up the test models for experiments. Although density of the Sawbones<sup>®</sup> blocks was also a sensitive parameter, it is not a parameter that we can control. Therefore, it was important to estimate how variations of Sawbones<sup>®</sup> properties may affect the test results. This motivated the following repeatability test.

#### 6.3.6 *Repeatability test.*

Results of the repeatability test are demonstrated in Figure 6.11. The low-density (Figure 6.11 [a]) and high-density blocks (see Figure 6.11 [b]) from three separate batches had a frequency variation of less than 50 Hz, implying that these blocks have very consistent material properties from batch to batch. In contrast, the hybrid blocks (see Figure 6.11 [c]) had a higher frequency variation up to 200 Hz, suggesting that material properties of hybrid blocks may vary from batch to batch considerably.



**Figure 6.11** Measured natural frequency values to detect variations of intact Sawbones<sup>®</sup> blocks: [a] low-density blocks (15-PCF), [b] high-density blocks (40-PCF), and [c] hybrid (40/20-PCF) blocks.

## 6.4 Discussion

In this chapter, our aim was to explore the vital importance of boundary conditions and experimental setup on the test results. Moreover, we evaluated the stability of Branemark dental implants embedded in Sawbones<sup>®</sup> of different densities by using the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, EMA, and FEA. Aside from the conclusive results provided above, there are several issues worth further discussion. They are:

1. combined use of EMA and FEA,
2. accuracy of the Osstell ISQ<sup>®</sup> measurements,
3. accuracy of the Periotest<sup>®</sup> measurements,
4. observations from EMA measurements, and
5. implications for the state of the art.

These issues are discussed in detail below.

#### 6.4.1 *Combined use of EMA and FEA.*

Through this study, we found that the combined use of EMA and FEA was an effective way to perfect test models and their setup. In general, natural frequency values are fairly robust quantities that can be predicted relatively accurately via FEA. Moreover, measured natural frequency values are sensitive quantities that can reliably reflect changes in the experimental setup. Therefore, the combined use of EMA and FEA can help us optimize the experimental setup. Here are some examples:

##### 6.4.1.1 Identification of residual stresses.

After some trial and error, the first successful version of our test models had an IMP abutment and side-anchorage as seen in Chapter 3. EMA measurements and FEA predictions corresponded closely. As we replaced the test models with a different abutment (LA), we noticed a considerable disagreement between the EMA measurements and the FEA predictions. The measured natural frequency values were not only inconsistent but also consistently higher than FEA predictions. The disagreement motivated us to further investigate the experimental setup. After some debugging efforts, we found that repeated clamping via the vise to retain the side-anchorage boundary condition had created permanent deformation of the Sawbones<sup>®</sup> blocks. This

is not surprising given the fact that Sawbones<sup>®</sup> are basically polyurethane foam blocks. The permanent deformation had induced residual stresses—stresses that remain in the object after the pressure applied to it or original cause is removed—and increased the natural frequency values. This is a critical finding, as it implies that the side-anchorage boundary condition is not reliable because the state of the residual stresses is always unknown. This was the major reason for us to investigate the bottom-positioned anchorage boundary condition as an alternative.

#### 6.4.1.2 Control of critical parameters.

After the bottom-positioned anchorage boundary condition was developed, we conducted a parametric study to identify critical parameters that needed to be strictly controlled in setting up the experiment. Results of the parametric study indicated that the contact area of the bottom-positioned anchorage (i.e., the area between the Sawbones<sup>®</sup> blocks and the metal base) was extremely critical. The effect could be in the order of 100-Hz change in the first natural frequency if the contact area is reduced by 10%. To securely control this parameter, we took several precautions in preparing the test models. First, we pressed the Sawbones<sup>®</sup> blocks firmly with hand pressure on the center of a metal block using carpet tape to obtain a large contact area. Second, we ensured that the tape was free of air bubbles to ensure a more solid contact area. Finally, we made sure that the Sawbones<sup>®</sup> blocks were fixed onto the metal block with no apparent gaps or rocking. After tightly controlling this parameter, we started to gain consistent test results.

#### 6.4.1.3 Use of repeatability tests.

After we controlled the experimental setup for bottom-positioned anchorage, we found that the measurements from the test models with high-density and low-density blocks were fairly consistent with the FEA predictions (e.g., within 20%). The measurements from the test models

with hybrid blocks, however, did not give comparable results with FEA. For example, let us consider the test model with LA and bottom-positioned anchorage shown in Figure 6.10 [c]. In this case, the FEA predicted the natural frequency values of the hybrid and low-density blocks to be the same or very close (see Figure 6.9 [c]), whereas the measured natural frequency values showed a 100 Hz difference (see Figure 6.10 [c]). Since all other parameters had been studied and tightly controlled, the only possible cause was variation of Sawbones<sup>®</sup> material. This led to the repeatability tests described in the section of “Materials and Methods.” The results of the repeatability tests (Figure 6.11) indicated that differences in the hybrid blocks led to large frequency variations from batch to batch. Therefore, we constructed a second test model with LA and bottom-positioned anchorage with hybrid blocks from a different batch. Measured natural frequency values of the second test model are shown in Figure 6.10 [d]. For the second test model, the natural frequency values of the hybrid and low-density blocks tended to be the same or very close as predicted in FEA. If the combined use of EMA and FEA had not been adopted, it would have been very difficult to spot the root cause of the abnormal frequency change shown in Figure 6.10 [c].

The measured results shown in Figure 6.10 [c] have significant consequences. As mentioned in preceding chapters, because the variations in measured natural frequency values is in the order of 50 Hz, one should be very cautious when using Sawbones<sup>®</sup>. The variations can be significantly larger when hybrid blocks are used. These variations need to be taken into account when interpreting test results.

The source of the frequency variations in Figures 6.10 [c] and [d] was worth further study. We weighed the Sawbones<sup>®</sup> blocks, measured their dimensions, and calculated their average

apparent density. Our calculations demonstrated that Sawbones<sup>®</sup> blocks from different batches may have a 3.7% difference in apparent density. According to the parametric study, a 3.7% difference in the average density should not result in a large variation in the natural frequency. The change of natural frequency values in Figure 6.10 [c] must have resulted from the stiffness effect (e.g., variations in Young's modulus).

Finally, we asked: Why do the hybrid blocks have a larger variation of natural frequency values from batch to batch than the high-density or the low-density blocks? One possible explanation is that the hybrid blocks are composite. They consist specifically of a 2-mm high-density laminate and a 40-mm low-density substrate glued together. The presence of a glue layer is well known to cause natural frequency values to change (Tsay et al., 1999). If the thickness and the stiffness of the glue layer could not be controlled precisely from batch to batch, the stiffness and natural frequency values of the entire hybrid block as a result would vary widely.

#### 6.4.2 *Accuracy of the Osstell ISQ<sup>®</sup> measurements.*

According to Haiat et al. (2014), implant length, implant positions, abutment length, bone quality and quantity, soft-tissue effects, and stiffness of the bone-to-implant interface are among the factors that are thought to affect ISQ readings. In the current study, all parameters but the implant length were fixed and firmly controlled during the experiments. With the combined use of FEA and EMA, we proved that the natural frequency values were independent of the implant length. Ideally, ISQ readings should reflect this observation and register no variation with respect to the implant length. The ISQ measurements, however, were not as satisfactory as hypothesized for several reasons. First, the ISQ measurements were somewhat scattered. For the high-density and hybrid blocks, the ISQ measurements were slightly scattered around a mean value that did not

seem to change significantly with respect to dental implant length. For the low-density blocks, the ISQ measurements were so scattered that they did not present any meaningful trend. Second, the ISQ measurements had an opposite trend to what was seen in the FEA and EMA. The ISQ values for the high-density and hybrid blocks were very close. In contrast, the FEA and EMA showed that natural frequency values for the high-density and hybrid blocks were significantly different (see Figures 6.9 and 6.10). Moreover, the measured ISQ values were higher for blocks with side-anchorage, while FEA and EMA showed higher natural frequency values for low-density blocks with bottom-positioned anchorage.

The Osstell ISQ<sup>®</sup>, as mentioned in preceding chapters, is the only current device available that can give relevant clinical information at various implant treatment phases. According to the literature, the Osstell ISQ<sup>®</sup> helps with evaluating implant primary stability and osseointegration, assessing feasibility of immediate loading, and monitoring bone density and implant stability over time (Haiat et al., 2014). As demonstrated in this study, however, the ISQ readings revealed the limitation of the use of the SmartPeg<sup>®</sup>, making the Osstell ISQ<sup>®</sup> measurements less useful in critical cases. The Osstell ISQ<sup>®</sup> values seem to depend greatly on the cortical layer or on the surface that the SmartPeg<sup>®</sup> is closer to or in contact with. Osstell ISQ<sup>®</sup> could not differentiate the trend among the three different density blocks.

The Osstell ISQ<sup>®</sup> findings in this dissertation actually resonate with many critiques published in the literature. For example, the Osstell ISQ<sup>®</sup> is thought to distinguish between implants with and without bone loss, but research does not support that idea (Chan et al., 2010). ISQ value is only somewhat reliable in assessing an integrated implant and the stiffness of the bone-to-implant interface (Nedir et al., 2004). In conditions where osseointegration is



questionable, the ISQ appears to fluctuate (Ersanli et al., 2005; Friberg et al., 1999; Krafft et al., 2015). It is also difficult to characterize a general standardized range of ISQ readings for a successful implant osseointegration for different implant systems (Ersanli et al., 2005; Krafft et al., 2015) and still the reliability of resonance frequency analysis (RFA) measurement remains low (Krafft et al., 2015; Lachmann et al., 2006). Therefore, the Osstell ISQ<sup>®</sup> cannot be used to provide definitive values for success, failure, or long-term prognosis of an implant (Haiat et al., 2014).

#### 6.4.3 Accuracy of the Periotest<sup>®</sup> measurements.

The Periotest<sup>®</sup> readings for the LA were much more consistent and easier to obtain than they were for the IMP abutment, particularly for the hybrid blocks. This might be attributed to several factors. First, LA has a much simpler geometry; it is a solid one-piece metal. In contrast, IMP abutment is a hollow, two-piece assembly adding more complications to the measurement. Second, LA is much shorter. LA is 7 mm long, whereas IMP abutment is 12 mm long. As a result, IMP abutment is not as stiff as LA. Therefore, a slight change in the measurement position or angle when using the Periotest<sup>®</sup> could lead to a larger scattering in measured PTV for IMP abutment compared to LA. For the high-density and hybrid blocks, the PTV tended to follow the FEA predictions and the EMA measurements. Basically, the bottom-positioned anchorage provided a stiffer boundary condition than side-anchorage, resulting in lower PTV readings (see Figures 6.6 [a] and [b]). Also, test models with LA were stiffer than those with IMP abutment, resulting in lower PTV readings; see Figures 6.6 [a] and [c]. However, we see that PTV is scattered, especially for the hybrid and low-density blocks even with a well-controlled experimental setup.

This observation of scattered PTV measurements also resonates with many critiques published in the literature. For example, research that has used the PTV to measure implant

stability reports low sensitivity of the device (Meredith et al., 1998) due to its wide value range (−8 to +50) (van Steenberghe et al., 1995). Moreover, PTV is strongly affected by the rod position along the height of the abutment and by direction (Derhami et al., 1995; Swain et al., 2008). Early studies reported that the Periotest<sup>®</sup> was a useful device for measuring implant stability (Kaneko, 1991; Lukas & Schulte, 1990; Manz et al., 1992a, 1992b; Olive & Aparicio, 1990; Schulte & Lukas, 1992; Winkler et al., 2001), but current reviews show some of the limitations and downsides of the instrument, such as the impact clinical variables have on the measurements, reduced resolution, low sensitivity, and susceptibility to operator variables when measuring dental implants (Aparicio et al., 2006; Atsumi et al., 2007; Faulkner et al., 2001; Hobkirk et al., 2006; Kaneko, 1994; Salvi & Lang, 2004; Swain et al., 2008). Part of the inadequacy in the Periotest<sup>®</sup> studies could be due to a lack of understanding of how the system being measured responds when excited. It is unclear how the Periotest<sup>®</sup> signals are affected by the interface properties and other physical properties of the measurement system.

From our experiment results, we observed that both the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV readings for the low-density blocks were more scattered, indicating that these devices have difficulty measuring implants placed in low-density blocks. Since implant failure is more likely to occur when the bone density is low, it is important that a testing device produces consistent readings, in this case to indicate impending failure. Therefore, the scattered readings of the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV for the low-density blocks show their inadequacy for detecting dental implant failure.

#### 6.4.4 *Observations from EMA measurements.*

The following is observed from the EMA natural frequency measurements. First, the measured natural frequency values were very consistent; with no scatter. Second, the natural frequency values did not vary as the implant length changed. Third, test blocks with LA were generally stiffer than those with IMP abutment, as manifested by a frequency increase in the order of 200 Hz; as demonstrated in Figures 6.10 [a] and [c]. Moreover, the boundary conditions of the test model played a very important role. Low-density blocks with bottom-positioned anchorage were stiffer than those with side-anchorage leading to higher natural frequency. With the bottom-positioned anchorage, the abutment-implant system tended to vibrate in a more constricted area. Finally, the EMA results correspond closely with the quantitative FEA predictions, with a margin of 20%.

#### 6.4.5 *Implications for the state of the art.*

Many studies have examined the effect of dental implant length on stability, and led to dissimilar results especially *in vitro*. Several studies have concluded that there is no effect of dental implant length on ISQ values or stability (Gomez-Polo et al., 2016; Meredith et al., 1996; Miyamoto et al., 2005; Monje et al., 2014). While some have shown an increase in ISQ with increasing implant length (Tsolaki et al., 2016), others have found a decrease in stability with increasing implant length (Ostman et al., 2006). We believe the reason for the conflicting results in these studies depends mainly on the experimental setup. For example, many of these studies (Ahn et al., 2012; Andres-Garcia et al., 2009; Bilhan et al., 2015; Brizuela-Velasco et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Lachmann et al., 2006; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al.,

2016; Sennerby et al., 2015; Tabassum et al., 2010; Tsolaki et al., 2016; Wang et al., 2015; Yamaguchi et al., 2015) place multiple implants in the same testing block, which we found to cause errors. Most studies used two or three implants that are very close in lengths (e.g. 10, 11 and 13mm), which does not give the full picture. (Gomez-Polo et al., 2016; Meredith et al., 1996; Monje et al., 2014).

Clinical studies have shown that implants short in length tend to have a lower survival rate (Chrcanovic et al., 2014; Winkler et al., 2000). They also have shown that implants longer in length last longer (Chrcanovic et al., 2014; Ferrigno et al., 2002). Nevertheless, implant length alone is not a definitive parameter for determining stability. The type of bone, position of the dental implant in the jawbone, and whether the patient is edentulous or dentate, as well as other factors collectively play a role in defining stability. In this chapter, we clearly demonstrated how bone density and the method of anchoring the implant-bone system can significantly affect the natural frequency values.

## **6.5 Conclusions**

Within the limitations of this study, we are reasonably able to conclude the following:

- As boundary conditions play an important role in dental implant testing, they should be taken into consideration when interpreting results of dental implant stability testing.
- The Osstell ISQ<sup>®</sup> data were somewhat scattered and unable to predict the trend in boundary conditions, whereas the Periotest<sup>®</sup> PTV data were scattered but nonetheless capable of accurately predicting trends.
- The combined use of FEA and EMA confirmed that:

- a. Test models with bottom-positioned anchorage yielded more consistent results than those with side-anchorage.
  - b. Test models with LA yielded higher frequency values than models with IMP.
  - c. Natural frequency values are not affected by changes in dental implant length.
  - d. Hybrid Sawbones<sup>®</sup> blocks tend to have large variations in measured natural frequency values, which may be due to the composite structure of these blocks.
  - e. A 10% change in Sawbones<sup>®</sup> block density significantly affected natural frequency values.
  - f. The EMA results quantitatively correspond closely with FEA predictions, with a 20% margin of error.
- The parametric study results confirm that the contact area of the bottom-positioned anchorage was the most critical parameter that needed to be highly controlled in setting up the test models for experiments.
  - Natural frequency values cannot be relied upon to accurately assess dental implant stability.

The next Chapter will describe a novel yet promising method for assessing the potential for dental implant stability.

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## **CHAPTER 7. A PROMISING NOVEL METHOD FOR QUANTIFYING DENTAL IMPLANT STABILITY *IN* *VITRO***

### **7.1 Introduction**

As discussed in previous chapters, there is great clinical need for a non-invasive device with a high level of sensitivity capable of detecting small changes in dental implant stability. A device proficient at accurately measuring smaller changes in dental implant stability would be helpful in many ways, including the need to: (a) evaluate dental implants at different stages of healing prior to the end result of successful osseointegration, (b) aid in critical diagnosis and treatment planning, (c) facilitate the decision process as to when an implant should be loaded, and (d) monitor implant status at the time of recall visits.

Some non-invasive devices currently used for assessing implant stability function on the basis of measurement of the resonance frequency of the implant-bone system (Atsumi et al., 2007; Buser et al., 2017; Westover et al., 2016), involve various disadvantages. First, these devices do not measure bone-to-implant interface stiffness, which is what determines dental implant stability

(Westover, et al. 2016). Second, these tests are often performed with the dental implant connected to an abutment or the testing instrument, which can significantly affect the measured frequency of the bone-implant system. Third, the measurement data can be scattered and inconclusive, for reasons relating to material property variations, insufficient joining of parts with abutment screws, and boundary conditions.

The preceding chapters confirmed that both the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> devices involve the three disadvantages described in the paragraph immediately above. Chapter 3 critically analyzed the operational principles of the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter in terms of theory. We proved that the resonance frequencies (in the case of the Osstell ISQ<sup>®</sup>), the contact time (for the Periotest<sup>®</sup>) and damping ratio (for the Perimeter) were not equivalent to bone-to-implant interfacial stiffness that is used to determine dental implant stability, thus confirming the first disadvantage. Next, we used experimental modal analysis (EMA) and finite elemental analysis (FEA) to calibrate the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter. We concluded that the presence of the abutment could significantly change the resonance frequencies of the implant-bone system, fully demonstrating the second disadvantage described in the paragraph above. Moreover, measurements from the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and Perimeter were in many cases scattered, pointing to the third disadvantage. Finally, in Chapter 6, we showed that variations of a few parameters (e.g., material properties and boundary conditions) could significantly affect measured resonance frequencies, posing further challenges for devices such as the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>.

To develop a non-invasive device that accurately measures dental implant stability, we must first figure out how to overcome the three disadvantages stated above. To overcome the first

disadvantage, an accurate mathematical model must be incorporated into the device's development. This mathematical model can subsequently be used to convert what is measured by the device (for example resonance frequency or contact time) to an assessment of stiffness at the bone-to-implant interface. To overcome the second disadvantage, we must be able to quantify the relationship between the bone-to-implant interfacial stiffness and the stability of the implant. For example, if two implants of different diameters and lengths exhibit the same degree of stiffness at the bone-to-implant interface, which implant is more stable? To be meaningful, the relationship must be defined in such a way that the quantified stability is independent of the abutment or anything connected to the implant. The accuracy of a measurement system involves two dimensions: (1) trueness—the closeness of the mean of the measurements to the actual value and (2) precision—the closeness of agreement within a set of results (ISO 5725-1) (ISO, 1998). To overcome the third disadvantage stated above, the device must adopt a reasonably reliable system of measurement for significantly reducing both noise and data scatter, thereby increased precision).

As mentioned in section 2.4.3 of Chapter 2, Westover et al. (2016) recently developed the Advanced System for Implant Stability Testing (ASIST), which offers a non-invasive means of arriving at a quantitative measurement of dental implant stability. The ASIST represents a modification of a Periotest<sup>®</sup> hand piece, to follow a 4-degree-of-freedom mathematical model. After the Periotest<sup>®</sup> hand piece measures the response during the contact, the linear stiffness at the bone-to-implant interface is estimated by means of the mathematical model. An ASIST stability coefficient (ASC) is derived on the basis of the estimated linear stiffness. The soundness of this methodology has been largely demonstrated through use of a model of bone-anchored hearing aid (BAHA) implants with abutments of various lengths. The results showed that the ASIST can not

only measure the degree of interfacial linear stiffness but also differentiate it from other system properties (Westover et al., 2016).

The work by Westover et al. (2016) is pivotal because it made significant progress for the first time in addressing the three disadvantages described above. It is not yet entirely clear, however, how effectively and accurately the interfacial stiffness data can be extracted when the ASIST is applied to dental implants in a porous substrate such as Sawbones<sup>®</sup>. It bears mention that BAHA-implants are significantly shorter than conventional endosseous dental implants. Questions remain as to the significant difference in bone-to-implant surface area, non-functional loading versus functional bite forces, and load bearing capacity in the context of several hundred pounds of chewing force. Clearly, several major challenges need to be addressed, as articulated below.

The first major challenge needing to be addressed is that the lumped-parameter, 4-degree-of-freedom model may be overly simplistic for use in measuring dental implant stability. The reason is that this model does not take into consideration various design features of dental implants, such as type of abutment, implant shape, and thread design. Moreover, accurate values of lumped parameters may be difficult to obtain, thus affecting the accuracy of the methodology. As a practical matter, what this means is that the stability measurement of two implant designs may not be evaluated on an equivalent basis.

The second major challenge with regard to the ASIST methodology is that the ASC may not be an entirely reliable way of quantifying stability. ASC is essentially comprised of the stiffness at the bone-to-implant interface (per unit area), multiplied by the diameter and the thread length, and then normalized with respect to the contact stiffness ( $K_I$ ) at the contact site. Here, there are two fundamental issues. Determination of the precise contact stiffness ( $K_I$ ) at the contact site is

problematic, thereby potentially affecting the stability readings. Moreover, the ASC by definition implies that an accurate measurement of linear stiffness in the direction of impact can reliably assess implant stability. Whether such a measurement can be used this way remains open to question, however.

The third major challenge with regard to the ASIST methodology is that the ASIST's experimental results may present a high degree of scatter. With the Periotest<sup>®</sup>, the relatively non-repeatable positioning and resulting non-ideal contact between the hand piece and the tooth/implant present a major source of data scatter (Faulkner et al., 2001; Hobkirk et al., 2006; Kaneko, 1994; Swain et al., 2008). Additionally, measurement noise cannot be averaged out during the process of ASIST measurements. These factors lead to less precise data (thereby increased scatter) from the ASIST methodology.

These three challenges were apparent in the BAHA-implant measurements (Westover et al., 2016). For example, Westover et al. (2016) found that samples with a higher ASC show slightly greater changes with variations in abutment length. When the degree of stiffness at the interface is high and the frequency values are also high, a slight alteration of the frequency response results in large changes in the ASC. In addition, the ASIST encountered problems when measuring different samples using the same implants in similar blocks. The ASC values varied greatly between the samples, to the point of being inconsistent.

The purpose of the study reported in this chapter was to develop and validate a novel system for estimation of dental implant stability. Experimental modal analysis (EMA) was used to obtain reliable measurements of natural frequency values and linear stiffness of a dental implant in an artificial bone substrate. Finite element analyses (FEA) were then performed, to model the

experimental setup and test environment. Finally, the combined use of EMA and FEA facilitated the estimation of angular stiffness of the dental implant in the Sawbones<sup>®</sup> substrate, serving as an index for quantifying the stability of a dental implant.

## **7.2 Materials and Methods**

### *7.2.1 Simulated jawbones.*

Sawbones<sup>®</sup> of three different densities were used: hybrid blocks (40/20-PCF), high-density blocks (40-PCF) and low-density blocks (15-PCF) as explained in Chapter 3, section 3.2.3.

### *7.2.2 Dental implants.*

Branemark Mk III implants in two different implant widths were used: (a) regular platform (RP), 4 mm in diameter by 8.5, 10, 11.5, 13, 15 and 18 mm length, and (b) wide platform (WP), 5 mm in diameter by 8.5, 11.5, 15 and 18 mm length.

RP implants of various lengths were positioned in the center of the three different-density Sawbones<sup>®</sup> blocks. Implants were placed following the manufacturer's surgical protocol and seated to 45 N·cm. Because WP implants are often used in low-density bone, the WP implants of different lengths were placed in 15-PCF Sawbones<sup>®</sup> blocks and seated to 45 N·cm. Locator abutments (LA) 7 mm in length were used for the RP implants, while LA 5mm in length were used for the WP implants. A 10 N·cm torque was used to secure the LA with the Branemark System Torque Control.

### *7.2.3 Boundary conditions.*

All samples were secured at the bottom of the block. The bottom surfaces of the Sawbones<sup>®</sup> samples were attached to a steel block (63.5 X 63.5 X 14.5 mm in dimension) using



carpet tape following the protocol used in Chapter 6 section 6.2.3. The steel block was clamped firmly in a vise. The goal was to create a fixed boundary condition at the bottom surface of the Sawbones<sup>®</sup> samples.

#### 7.2.4 *Measurement methods.*

The implants were tested using EMA. The experimental setup is described in detail in Chapter 3, section 3.2.5. Essentially, the setup consisted of a force hammer, a laser Doppler vibrometer (LDV) and a spectrum analyzer. The hammer measured the force applied to the implant, and the LDV measured the velocity response of the implant. Based on the measured force and velocity, the spectrum analyzer calculated frequency response functions (FRF) from which natural frequency values of the implant-Sawbones<sup>®</sup> test models were extracted. The average stiffness of implant-abutment-Sawbones<sup>®</sup> for each sample was extracted from the FRF as follows.

##### 7.2.4.1 Stiffness coefficients.

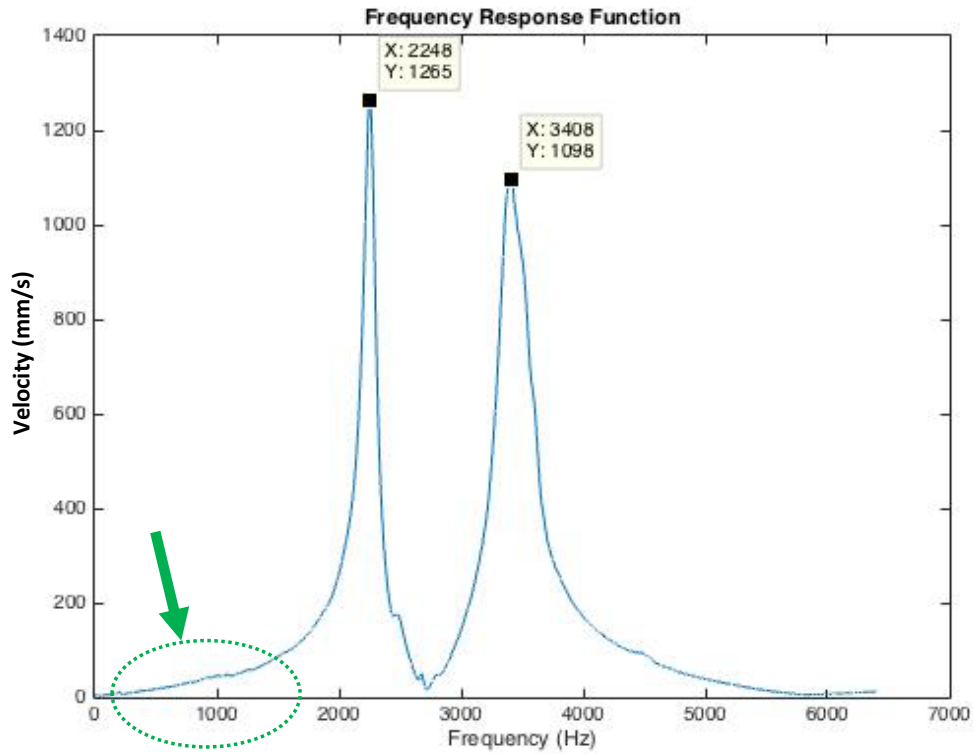
Stiffness is the rigidity of an object. It describes the relationship between an applied load on a structure and the responding displacement. Stiffness depends on the location and type of the load as well as the displacement. Two common forms of stiffness are [a] linear stiffness and [b] angular stiffness.

**[a] Linear stiffness coefficient** For linear stiffness coefficient, the load is a force  $F$  and the structural response is a linear displacement  $x$ . The applied force is proportional to the displacement, and the ratio is the linear stiffness coefficient  $k$ , i.e.,

$$k = \frac{F}{x}$$

The linear stiffness coefficient  $k$  can be estimated in the frequency domain. Theoretically,

the FRF (with force input and velocity output) should increase linearly when the frequency is significantly less than the first natural frequency. This is the area of the graph (flat region) before the first peak (i.e., in the frequency range before it hits the first peak) (see Figure 7.1). Moreover, the slope of the FRF is the reciprocal of the linear stiffness coefficient  $k$  of the measured implant-bone-abutment assembly. This is, however, a somewhat less robust measurement (compared with natural frequency values) because the measured FRF may not truly have a linearly increasing region due to measurement errors. In other words, the area before the first peak may not always be linear, making it difficult to estimate the linear stiffness. In addition, the linear stiffness coefficient  $k$  estimated depends on where the hammer strikes and where the laser is directed, which may vary between specimens. Nevertheless, it is a measurable quantity that should be interpreted carefully.



**Figure 7.1** Linear stiffness coefficient estimation via the linear region of the frequency response function before the first frequency peak. The linear stiffness coefficient is the reciprocal of the slope of the linear region.

**[b] Angular stiffness coefficient:** For angular stiffness coefficient, the load is an applied moment  $M$  and the structural response is an angular displacement  $\theta$ . The applied moment is proportional to the angular displacement, and the ratio is the angular stiffness coefficient  $k_\theta$ , i.e.,

$$k_\theta = \frac{M}{\theta}$$

The angular stiffness coefficient is an ideal representation of dental implant stability. In theory, when a moment is applied at the base of an abutment, the moment does not deform the abutment elastically. The response is entirely from the bone-to-implant interfacial stiffness and the elasticity of the implant. The angular stiffness coefficient is an excellent way to quantify dental

implant stability, because it completely removes the effects of the abutment.

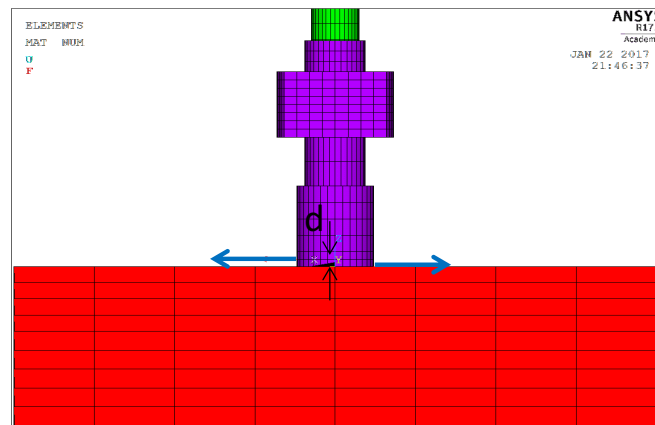
The angular stiffness coefficient  $k_\theta$ , however, is quite difficult to measure experimentally. Application of torque to the implant without incurring any abutment deformation is unattainable. Angular displacement is also difficult to measure accurately. In this chapter, we describe a combined approach to estimate the angular stiffness coefficient  $k_\theta$ . We first measured natural frequency values via experiments. Then we relied on an accurate mathematical model (i.e. via FEA) to convert the measured natural frequency values to estimate the angular stiffness coefficient  $k_\theta$ .

#### 7.2.5 Finite element analyses (FEA).

A 3D finite element model (FEM) was created using ANSYS R-15 to simulate the Branemark implants, the experimental setup, and the test results. Material properties of each component (e.g., Sawbones<sup>®</sup> and implants) as published by manufacturers were used in the FEM (see Table 3.1). The model is explained in detail in Chapter 6. Two implant widths were modelled – RP and WP. For both implants, the model included two types of abutments: locator abutment (LA) (7 mm in length for RP and 5mm in length for WP) and impression coping abutment (IMP, 12 mm in length). In addition, the LA was used to verify the experimental measurements, whereas the IMP abutment was used to demonstrate the validity of quantifying implant stability via the angular stiffness coefficient  $k_\theta$ .

After the model was created, a modal analysis was conducted to calculate the natural frequency values. Then a static analysis was performed to calculate linear and angular stiffness coefficients  $k$  and  $k_\theta$ . To calculate the linear stiffness coefficient  $k$ , a force was applied to a node where in the experiment the impact hammer contacted. The displacement of a node where the LDV

measured experimentally was then predicted via FEA. The ratio of the force and the displacement predicted the linear stiffness coefficient  $k$ . To calculate the angular stiffness coefficient  $k_\theta$ , a pair of equal and opposite forces were applied to two neighboring nodes at the base of the abutment to form a couple (see Figure 7.2). The rotation of the centerline of the abutment at its base was the angular displacement. The ratio between the couple and the angular displacement gave the angular stiffness coefficient  $k_\theta$ .



**Figure 7.2** To estimate the angular stiffness coefficient  $k_\theta$ , a pair of equal and opposite moments applied to two neighboring nodes at the base of the abutment, forming a couple, in order to calculate the angular stiffness coefficient ( $k_\theta$ ). The rotation of the centerline of the abutment at its base was the angular displacement, and the ratio between the couple and the angular displacement yielded the angular stiffness coefficient  $k_\theta$ .

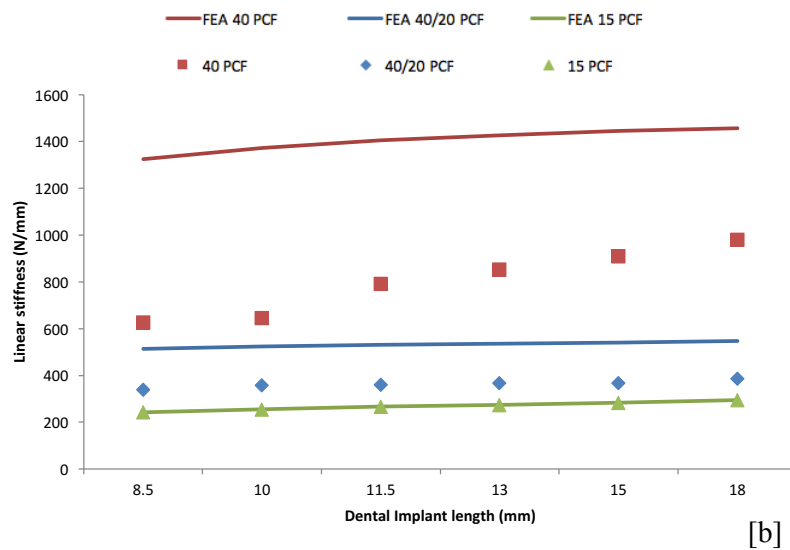
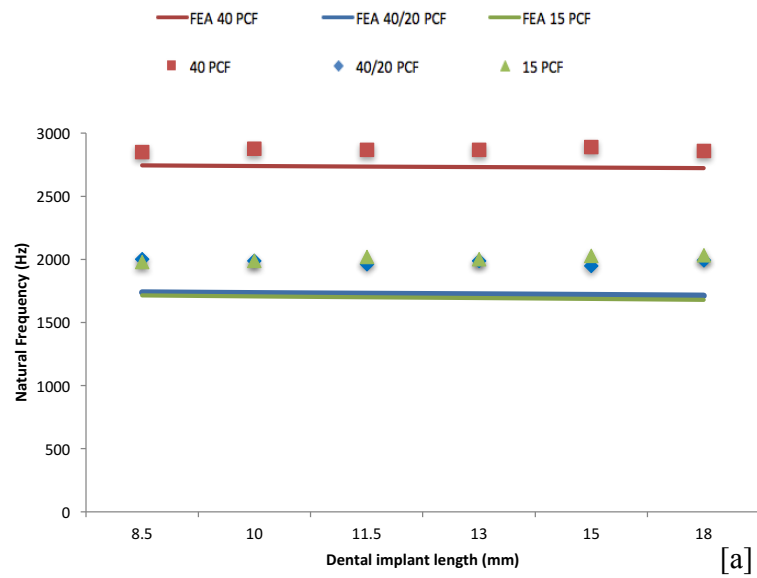
## 7.3 Results

### 7.3.1 *Results for Branemark RP implants.*

#### 7.3.1.1 Experimental modal analysis (EMA).

For RP implants with LA, Figure 7.3 [a] shows the measured first natural frequency values versus the implant length. The measured natural frequency values fell in the range between 2848 and 2888 Hz for the high-density blocks, 2112 and 2176 Hz for the hybrid blocks, and 1936 and 2036 Hz for the low-density blocks. The measured natural frequency values did not have a significant correlation to the dental implant length.

Figure 7.3 [b] shows the linear stiffness coefficients extracted from the measured FRF. The estimated linear stiffness coefficients for the high-density blocks ranged from 628.6 to 980.1 N/mm for the high-density blocks, from 329.5 to 386.7 N/mm for the hybrid blocks, and from 169.8 to 271.5 N/mm for the low-density blocks.

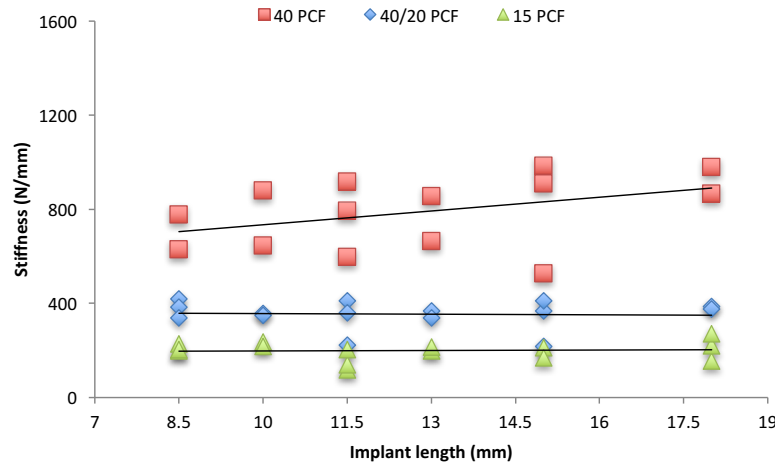


**Figure 7.3** Measurements from experimental modal analysis (EMA) for RP implants with locator abutment (LA) and their comparison with finite element predictions: [a] the first natural frequency values vs. implant length, and [b] linear stiffness vs. implant length. (Markers and solid lines represent EMA measurements and FEA predictions, respectively.)

In general, the stiffness had a similar trend to the natural frequency values, where the high-density blocks were significantly higher in stiffness and natural frequency values compared to those of the hybrid and low-density blocks. The low-density and hybrid blocks had closer linear stiffness coefficients and natural frequency values. The linear stiffness coefficient values for the high-density blocks increased with increasing implant length, while the linear stiffness coefficient values for the hybrid and low-density blocks did not vary significantly with dental implant length.

To evaluate the repeatability and consistency of the estimated linear stiffness coefficients, the experiments were repeated. FRF results from multiple EMAs were used to estimate the average linear stiffness. The linear stiffness coefficient was estimated from FRFs of the Branemark RP implants that had consistent natural frequency measurements. The results are demonstrated in Figure 7.4. The linear stiffness coefficients varied widely especially for the high-density blocks, but fell in the same range for the hybrid and low-density blocks. Two important points need to be emphasized. First, this shows how linear stiffness coefficient can be a hard parameter to estimate since it depends on where the hammer taps and the laser is aimed. This can vary from one test to another regardless of how much the one attempts to control the parameters. Second, detecting the low-stiffness cases is critical clinically since these are the conditions in which implants are likely to fail.





**Figure 7.4** Repeatability test results showing variations of extracted linear stiffness coefficients.

#### 7.3.1.2 Finite element analysis (FEA).

**[a] Natural frequency values.** Natural frequency values predicted by FEA are presented in Figure 7.3 [a] for comparison. The three solid lines represent the FEA predictions of natural frequency values for hybrid, high-density, and low-density blocks. In general, the predicted natural frequency values correspond closely, with the measured natural frequency values within 10%. The high-density blocks showed better agreement than the low-density and hybrid blocks. Moreover, the FEA predictions and the EMA measurements had the same trend. For example, natural frequency values showed no correlation to the implant length. Natural frequency values of the hybrid blocks and low-density blocks were also about the same.

Therefore, the close agreement between the FEA predictions and EMA measurements in natural frequency values indicates that the FEM is very accurate. Note that the FEA predictions and EMA measurements were not the same due to variables in the experiments such as differences

in the material properties of Sawbones<sup>®</sup> and contact conditions at the implant-Sawbones<sup>®</sup> interface (see Figure 7.3[a]).

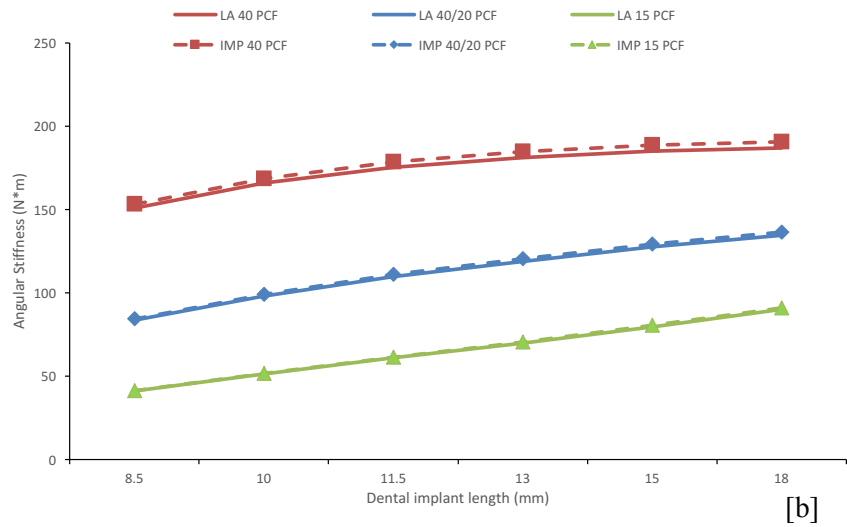
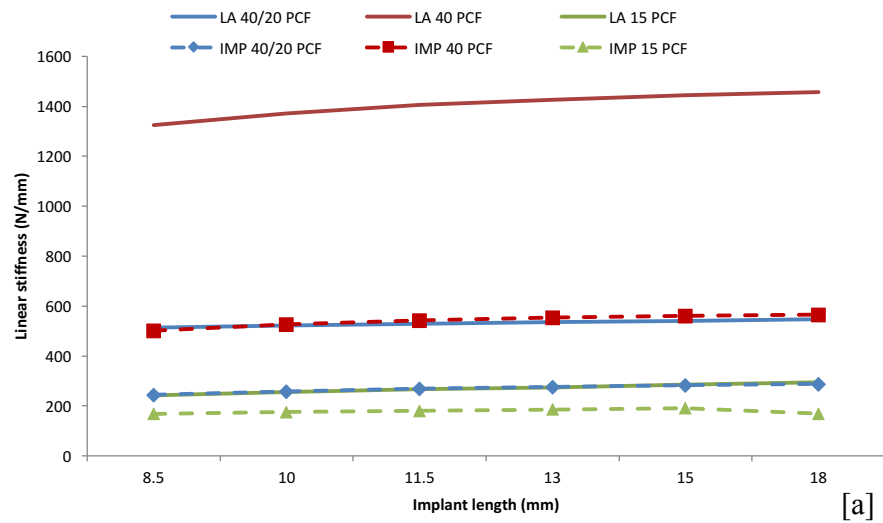
**[b] Linear stiffness coefficient.** Linear stiffness coefficients predicted by FEA are shown in Figure 7.3 [b] for comparison. There are several things to note. First, the FEA predictions had the same trend as the EMA results. Specifically, the linear stiffness coefficients of the high-density blocks tended to increase with increasing implant length. In contrast, the linear stiffness coefficients of the low-density and hybrid blocks were closer to each other, and did not vary significantly with respect to dental implant length. The FEA predictions captured all the features observed in the EMA measurements. Second, the predicted linear stiffness coefficients were higher than the measured ones. For the low-density Sawbones<sup>®</sup>, the predicted and measured linear stiffness coefficients were close. For the high-density Sawbones<sup>®</sup>, the difference between the predicted and measured linear stiffness coefficients was large but had the same trend.

The results shown in Figure 7.3 [b] are not surprising; instead, they are encouraging. As explained earlier, it is well known that linear stiffness coefficient is a less robust quantity than natural frequency values measured by EMA. Linear stiffness coefficient is very susceptible to test conditions (e.g., location of the hammer taps), and has a much smaller signal-to-noise ratio compared with natural frequency measurements. Therefore, it is not realistic to expect that numerical values of the linear stiffness coefficients from the FEA predictions and EMA measurements would correspond closely. Nevertheless, the comparison in Figure 7.3 [b] shows two encouraging signs. First, the FEA results predicted the trend. That again indicates that the FEM is accurate. Second, the biggest difference between the predicted and measured linear stiffness coefficients was around 100% (e.g., 1300 vs. 620 N/nm for 8.5 mm implant in high-

density block). This difference— between the predicted and measured linear stiffness coefficients—is reasonable considering the fact that an impact hammer was used to assess stiffness.

The FEM also proves that linear stiffness coefficients are not good indicators of implant stability, because they depend heavily on abutment geometry. Figure 7.5 [a] compares the FEA predictions of the linear stiffness coefficients with an LA and an IMP abutment. The change from LA to IMP abutment did not change the trend of the linear stiffness coefficients with respect to the implant length. The linear stiffness coefficients with the IMP abutment, however, were significantly lower. This indicates that the elasticity of the IMP abutment affected the linear stiffness. Therefore, linear stiffness is not a representative measure of the bone-to-implant interface stiffness or the stability of the implant.

**[c] Angular stiffness.** The FEM allows us to calculate angular stiffness coefficients. Figure 7.5 [b] compares angular stiffness coefficients predicted by FEA for models with LA and IMP abutments. When a moment was applied at the cervical area of the dental implant or the base of the abutments (LA and IMP) with Branemark RP in hybrid blocks, we noticed that the angular stiffness coefficients were quite similar with minimal difference for implants with both abutments – less than 1.5%. This shows how the angular stiffness coefficient at the base or apical region of the abutment is independent of the length of the abutment and can represent the dental implant stability more accurately. The angular stiffness coefficients tended to increase significantly with increasing dental implant length as demonstrated in Figure 7.5 [b].



**Figure 7.5** FEA predictions of stiffness in RP implants with LA and IMP abutments: [a] comparison of linear stiffness coefficients, and [b] comparison of angular stiffness coefficients.

### 7.3.2 *Results for Branemark WP implants.*

The EMA and FEA procedure developed above can be applied to WP implants to determine their stability. Moreover, the stability can be compared with that of the RP implants for an evaluation. Note that the WP implants were measured only in low-density blocks. Therefore, the comparison is made with RP implants in low-density blocks.

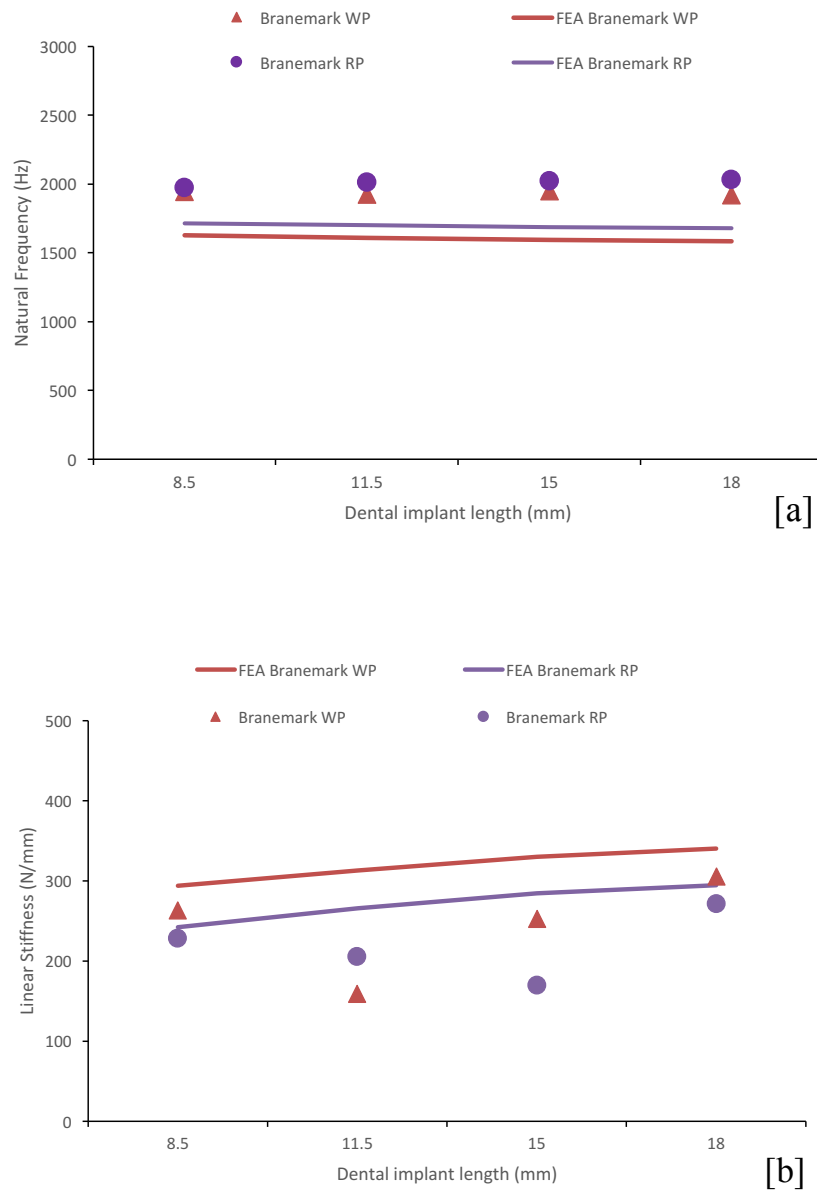
#### 7.3.2.1 Experimental modal analysis (EMA).

The first natural frequency values measured via EMA ranged from 1984 to 2072 Hz; see Figure 7.6 [a]. The natural frequency values were in the same range as those for the RP implants in low-density blocks. Therefore, varying the diameter does not cause a significant change in natural frequency values. In contrast, the linear stiffness coefficients varied from 346.7 to 429.6 N/mm, which is significantly higher than those extracted from the RP implants in low-density blocks; see Figure 7.6 [b]. It is evident that natural frequency values are not representative of implant stability.

#### 7.3.2.2 Finite element analysis (FEA).

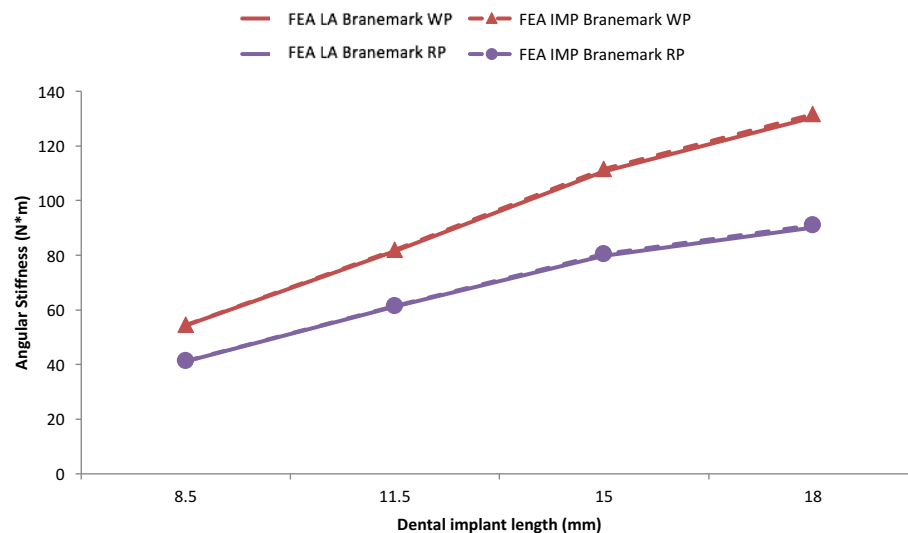
**[a] Natural frequency values.** FEA was conducted on WP implants with LA in low-density blocks. Natural frequency values predicted by the FEA are presented in Figure 7.6 [a] for comparison. The predicted natural frequency values corresponded closely with the measured ones.

**[b] Linear stiffness coefficient.** The linear stiffness coefficients predicted by FEA for WP and RP implants are shown in Figure 7.6 [b] for comparison. Note that the predicted linear stiffness coefficients corresponded closely with the measured stiffness not only in magnitude but also in trend. For example, the predicted stiffness for WP was higher than that of RP.



**Figure 7.6** Comparing the RP and WP implants in low-density blocks measured by EMA and predicted by FEA: [a] the first natural frequency values, and [b] linear stiffness coefficients.

**[c] Angular stiffness coefficient.** The angular stiffness coefficient was predicted for Branemark WP and RP implants in low-density blocks; see Figure 7.7. Both angular stiffness coefficients obtained from models with LA and IMP abutments are also presented in Figure 7.7. There are several observations to make: First, the angular stiffness coefficients from the LA and IMP abutments were almost identical, proving again that the angular stiffness coefficient is a robust method to assess implant stability. Second, WP implants had significantly larger angular stiffness coefficients than the RP implants. Finally, the angular stiffness coefficients increased with increasing implant length for both WP and RP implants.



**Figure 7.7** FEA prediction of angular stiffness in RP and WP implants in low-density blocks (15-PCF), with moment applied at the bottom of the abutment. The angular stiffness coefficient is almost identical for FEA models with LA or IMP abutments.

## 7.4 Discussion

Through the experimental and numerical results presented above, there are several issues that warrant in-depth discussion. One is the quantification of implant stability. The second issue is the accuracy of the presented new approach. Another issue is the possible modification for field applications. These are discussed in detail below.

### 7.4.1 *Quantification of dental implant stability.*

The results obtained from the RP and WP implants illustrated that the angular stiffness coefficient is an effective measure to quantify implant stability. This is both easy to understand and intuitive; in practice, when a clinician taps on a dental implant, he or she is getting a sense of the angular stiffness. When a person jiggles his or her own teeth, the person is estimating the stability. The angular stiffness coefficient is theoretically sound because it is almost independent of the size of the abutments. Another advantage of using the angular stiffness coefficient to measure implant stability is that it is not a normalized quantity. That means one can predict how much the implant will displace under various biting conditions. Moreover, implants of different designs or implants in different bone densities, for example, can be compared and differentiated on an equal footing.

The angular stiffness coefficient, however, is difficult to measure directly in experiments. If natural frequency values or linear stiffness coefficients are measured experimentally, the angular stiffness coefficient may be extracted via an accurate mathematical model. More research should be directed into these areas to make measurements or extraction of angular stiffness coefficients more accurate and expedient.



Traditionally, natural frequency values (or resonance frequencies) and contact time have been used as a measure of implant stability, as in the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>, respectively. As discussed in Chapter 3, natural frequency values and contact time do not entirely represent implant stability in theory. In this study, we confirmed that the concept of using natural frequency values to quantify implant stability is inaccurate and obsolete. For example, RP implants of different lengths had roughly the same natural frequency values, as seen in Figure 7.3 [a]. Nevertheless, RP implants with a longer implant length had better implant stability because of their higher angular stiffness (see Figure 7.5 [b]). Similarly, WP and RP implants had similar natural frequency values (see Figure 7.6 [a]), but had significantly different angular stiffness coefficients (see Figure 7.7). The use of natural frequency values cannot determine implant stability.

#### *7.4.2 Accuracy of the new approach (using angular stiffness coefficient to measure implant stability).*

Since the goal is to quantify dental implant stability, it is extremely important to secure accurate values of angular stiffness coefficients. This depends on two factors: reliable experimental measurements and an accurate mathematical model. In this study, natural frequency values manifested themselves as very reliable experimental measurements. They were robust and repeatable. In contrast, linear stiffness coefficients were not. They were very sensitive to experimental setup, and the measured linear stiffness coefficient values presented some degree of scatter (see Figure 7.4). Therefore, the measured natural frequency values, rather than the measured linear stiffness coefficients, should be used for finding the angular stiffness coefficient.

In this study, the FEM proved to be very accurate; the model predictions were in agreement with the measured natural frequency values. The material properties data from manufacturers was used in the FEM (e.g., density and Young's modulus of Sawbones<sup>®</sup>) (Table 3.1). These material properties must be used to calibrate the FEM against experimental measurements to prove its accuracy.

The measured natural frequency values were proven to be reliable and the FEM was accurate. One can confidently conclude that the angular stiffness coefficients presented in Figures 7.5 and 7.7 accurately estimate the true angular stiffness coefficients of the experimental setup. Thus, it can be concluded that WP implants have higher stability than RP implants. In addition, stability is positively correlated with dental implant length, where implants with longer length have higher stability than those with shorter length.

It is also important to interpret the measured linear stiffness coefficients accurately. In this study the predicted and measured linear stiffness coefficients corresponded closely for WP and RP implants in low-density (15-PCF Sawbones<sup>®</sup>) blocks but not as well for several other cases (e.g., RP implants in high-density [40-PCF Sawbones<sup>®</sup>] blocks). Since the linear stiffness coefficient is a less robust quantity to measure in experiments, existence of some disagreement does not necessarily mean that the FEM for RP implants in 40-PCF Sawbones<sup>®</sup> was less accurate. In contrast, the FEM was proven accurate because it predicted the natural frequency values very well. The measurements of the linear stiffness coefficients only imply that the linear stiffness measurements are more reliable for WP and RP implants in 15-PCF Sawbones<sup>®</sup> blocks than for RP implants in 40-PCF Sawbones<sup>®</sup> blocks. Moreover, when FEA predicted and EMA measured

linear stiffness coefficients agree with each other, it provides further assurance of the accuracy of the FEM.

#### *7.4.3 Modifications for field applications.*

The purpose of this study was to prove the concept that the angular stiffness coefficient is a viable and reliable method for measuring dental implant stability. All parameters in the experiments were controlled as much as possible. The geometry of test blocks, materials used, and boundary conditions were well controlled. There were some uncontrollable parameters, however, such as the material properties of the Sawbones<sup>®</sup> blocks used in the experiments. Sawbones<sup>®</sup> materials are indeed nonlinear; they have different Young's modulus in tension and in compression. The force-displacement curves of Sawbones<sup>®</sup> materials are not entirely linear. It is also not clear how homogeneous Sawbones<sup>®</sup> materials are and it is unknown if the Young's modulus is uniform throughout the entire block. These uncontrollable parameters potentially led to the differences between the measured and predicted natural frequency values.

In practical application, mechanical properties of the bone surrounding the dental implant is unknown. Therefore, the appropriate practice is to adjust the Young's modulus of the substrate surrounding the implant so that the natural frequency values predicted by FEA match with the EMA measured ones, meaning that one needs to fit the stiffness of the substrate materials from the experimental measurements. This is exactly what Westover et al. (2016) did in their work. The fitted substrate stiffness together with the FEM produces an angular stiffness coefficient that better reflects the experimental setup.

Note that natural frequency values are used to match the FEA predictions and EMA measurements because they are very robust in experimental measurements. If the linear stiffness

coefficients can be measured with equal or better reliability and robustness, one can adjust the Young's modulus of the substrate surrounding the implant so that the FEA predicted linear stiffness coefficients match with the EMA measured values.

## **7.5 Conclusions**

Within the limitations of this study, we are reasonably able to conclude the following:

- Natural frequency values do not accurately represent dental implant stability. Changes to implant length and diameter did not affect natural frequency values.
- Linear stiffness coefficient did not prove to be a good measure of implant stability, as this quality was significantly affected by the presence of abutments. It was also very sensitive to any changes in the testing environment. Discrepancies of 50-75% were observed between EMA and FEA predictions when using the high-density (40-PCF) Sawbones<sup>®</sup> blocks for measuring linear stiffness coefficient.
- Angular stiffness coefficient can be measured through the combined use of experimental measurements and mathematical modeling. For example, the natural frequency values of a dental implant can be extracted experimentally via EMA, after which a reliable mathematical model (e.g., a finite element model) can relate the natural frequency values to the degree of angular stiffness. The combined use of EMA and FEA to estimate angular stiffness is demonstrated in two examples in this chapter: (a) RP implants of different lengths for three different Sawbones<sup>®</sup> densities, and (b) RP and WP implants with low-density (15-PCF) Sawbones<sup>®</sup> blocks.
- Although the angular stiffness coefficient is independent of the type of abutment present, it is difficult to measure directly in experiments.

- The angular stiffness coefficient represents a much more reliable measurement for quantifying dental implant stability. Unlike other measurements, angular stiffness coefficient was a much more sensitive measurement of implant stability, demonstrating that wider implants had higher stability than narrower implants. Similarly, implants longer in length exhibited greater stability than those of shorter length.

The following, final, chapter presents a summary of this dissertation's conclusions, as well as some ideas for future research.

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## **CHAPTER 8. CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE RESEARCH**

The main objective of this dissertation was to define a parameter capable of measuring dental implant stability, a more precise and reliable parameter for detecting failing dental implants than is currently in standard use. Through a series of studies presented in the preceding chapters, we successfully uncovered a promising novel method for assessing dental implant stability.

This final chapter will summarize the dissertation, reviews its findings and limitations, discuss its contributions, and also outline recommendations for future research.

### **8.1 Summary of Dissertation Findings**

During the past decade in particular, the use of dental implants to replace missing teeth has increased dramatically. The overall high success rate notwithstanding, any significant percentage of failure is worrisome. Dental practitioners are capable of identifying implants that are successful,

as well as those that have obviously failed. Detecting an implant before it fails, however, remains a major issue. Dental practitioners have traditionally relied upon percussion and instrument tests to evaluate dental implant stability. Radiographs are routinely utilized for evaluating conditions relating to the critical aspect of bone-to-implant interface. Together, these serve as the most commonly employed diagnostic tools for clinicians. Experimental tests—i.e., the reverse torque test, histomorphometric analysis, implant pull-out, and push-in tests—utilize measures that are more objective and quantifiable than the clinical techniques mentioned above, but are more destructive and therefore used mainly in experimental studies (Atsumi et al., 2007). Few of the many non-invasive devices proposed for measuring dental implant stability are actually available at present for clinical use, the way that the Osstell ISQ® and the Periotest® are. Challenges in this realm are further aggravated by the fact that current measurement technologies are not well referenced, in the sense that it is not apparent how a particular number that results from a particular measurement relates to the numbers in other systems. Other issues with current measurement technologies are that they are not highly precise, have low levels of discrimination, and have little documented correlation to implant health. Despite the importance of the need for measuring dental implant stability, an effective, reliable methodology has eluded researchers, with current methods used to detect implant stability either in the early stages or incapable of providing a complete picture with regard to stability.

Although many published studies have been devoted to assessing dental implant stability, and numerous publications have sought to identify the biomechanical factors involved (Bilhan et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016;



Sennerby et al., 2015; Tabassum et al., 2010; Wang et al., 2015; Yamaguchi et al., 2015) a close examination of these studies reveals a number of significant flaws, regarding both methodologies and results. *In vitro* test models used in these studies have for the most part consisted of dental implants placed in Sawbones<sup>®</sup> (Ahmad & Kelly, 2013; Ahn et al., 2012; Giovanni et al., 2011; Hsu et al., 2013; Tabassum et al., 2010; Wang et al., 2015), which can be explained by the fact that no standard protocol is available for clinical use and the placement of dental implants in Sawbones<sup>®</sup> presents itself as the easiest option. Previous studies on dental implant stability (Ahn et al., 2012; Bilhan et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Mazzo et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016; Sennerby et al., 2015; Tabassum et al., 2010; Wang et al., 2015; Yamaguchi et al., 2015), have not devoted sufficient attention to the experimental setup of the test models, thereby calling into question the reliability of their results. Would it be better to place implants in single blocks or all in one block? What would be the most effective way to clamp these blocks, for purposes of stabilizing them? What would be the most appropriate size or dimensions for each Sawbones<sup>®</sup> block? These are all examples of key questions that deserve to be considered in the experimental setup, questions that previous studies did not squarely address, pointing to the critical need for a standard protocol for the use of synthetic blocks (Sawbones<sup>®</sup>) in implant stability testing.

A non-invasive device with a high level of sensitivity, capable of detecting small changes in dental implant stability, would be helpful in many ways, including the need to: (a) evaluate dental implants at different stages of healing, prior to the desired end result of successful osseointegration, (b) aid in the planning of critical diagnosis and treatment, (c) facilitate the

decision process as to when an implant should be loaded (made functional), and (d) monitor implant status at the time of recall visits.

In this study, we sought to: (1) identify the key biomechanical factors involved in dental implant stability, (2) understand the mechanics by which current dental implant stability devices work, (3) develop a new method that is capable of efficiently identifying the key biomechanical factors involved in dental implant stability, a method that is also capable of measuring and assessing the stability of dental implants with greater accuracy and reliability. In the course of realizing these objectives, we arrived at a standard protocol for the use of synthetic blocks (Sawbones<sup>®</sup>) in implant stability testing.

In the process of achieving the objectives described above, several fundamental questions regarding implant stability and measurements were addressed:

- What is the most clinically meaningful way to define implant stability?
- To what extent is dental implant stability affected by either bone density or the length and diameter of the implant?
- How accurate and consistent are the implant stability measurement devices currently in use—and what properties are they actually measuring?

These questions were answered in the studies described in the preceding chapters, as follows:

In Chapter 3, we evaluated the accuracy and reliability of the two commercial devices currently available for testing implant stability: the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>. We also evaluated the Quantitative Percussion Diagnostics device (aka the Perimeter), which at the time of the study was being tested in prototype form. These devices (the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter) were evaluated by calibrating them using both experimental modal analysis

(EMA) and finite element analysis (FEA), on the basis that EMA and FEA are known to be proven experimental and numerical (theoretical) methods, respectively, for measuring natural frequency values, viscous damping factors, mode shapes, and stiffness of a mechanical structure. To facilitate the calibration, we placed dental implants of different lengths in synthetic bone models of different densities, examining how length affects implant stability. Specifically, measurements taken by the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter were compared with those obtained by means of both EMA and FEA. Second, we analyzed the theoretical mechanics on which the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup> and the Perimeter are based.

From this study, we found that the combined use of EMA and FEA serves as a rigorous and reliable means for calibrating commercial implant stability devices. Results from both EMA and FEA were consistent with each other not only qualitatively but also quantitatively. Both EMA and FEA demonstrated that natural frequency values are not affected by implant length. Numerical values of natural frequency values obtained from EMA were within 20% of FEA predictions.

Successful development of an experimental model for testing implant stability represented the most challenging part of this dissertation study, largely because of the absence of guidelines available for testing dental implant stability *in vitro* in general, and specifically in Sawbones<sup>®</sup>. Through the combined use of both EMA and FEA, we came to understand that several factors are necessary to appropriately set up a test model. First, it is not desirable to have multiple implants placed in a large Sawbones<sup>®</sup> block, as each implant vibration is affected by the presence of neighboring implants. Instead, each implant should be placed in its own test block, preferably in the center, giving all implants the same setup. Second, dimensions of the test blocks must be the same and the orientation of the abutment must also be the same relative to the Sawbones<sup>®</sup> blocks.

Third, because the boundary conditions stabilizing the Sawbones<sup>®</sup> blocks are critical, they must be maintained in the same manner for all Sawbones<sup>®</sup> blocks during the tests. The combined use of EMA and FEA allowed us to identify sensitive parameters that needed to be well-controlled in our test models, thereby substantially improving our experimental models.

In addition, the test results from the Osstell ISQ<sup>®</sup> were partially consistent with the benchmark (EMA and FEA results), whereas the test results from the Periotest<sup>®</sup> deviated wildly. Moreover, results from the Perimeter and the damping ratio estimation did not agree. The inconsistency of the test results from the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter was explained by analyzing the mechanical theories, physics and equations of these stability devices. We observed several fundamental issues underlying the concept of non-invasive testing for dental implant stability. The logic of these devices (the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter) rests on the basic assumption of a single-degree-of-freedom system. For example, when a Periotest<sup>®</sup> or an Osstell ISQ<sup>®</sup> measurement is performed, a number (e.g., ISQ or PTV) is obtained. If the assumption is not met, the number obtained would not be meaningful. Currently, there is no systematic way to check whether the single-degree-of-freedom assumption is met. This poses a real problem with the use all these non-invasive testing devices, because it is not clear which measurements are meaningful and which should be ignored. In addition, none of these devices directly measures implant stability. The Perimeter hypothesizes that damping is correlated with implant stability. Therefore, damping properties are measured. The damping ratio, however, is a difficult parameter to quantify, potentially involving a significant degree of error. The Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> both hypothesize that generalized stiffness is correlated with implant stability—i.e., the greater the stiffness, the better the stability. Generalized stiffness, however, may depend

on many parameters, such as bone density, implant length, threads, and healing conditions at the implant-bone interface. To obtain the generalized stiffness, the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> measure natural frequency and length of contact time, respectively. Extracting the generalized stiffness accurately from the measured natural frequency and length of contact time, however, is not an easy matter. It is not clear whether algorithms embedded in the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> convert the natural frequency and the contact time into generalized stiffness with a reasonable degree of fidelity.

We realized, furthermore, that natural frequency values do not represent implant stability, as natural frequency values are “system-level” quantities. In other words, natural frequency values are properties of the entire implant-bone-abutment system, as opposed to properties relating exclusively to the bone-to-implant interface (e.g., stiffness at the interface). Since the natural frequency is a system-level quantity, the degree of generalized stiffness derived from a measured natural frequency would also be a system-level measurement.

As mentioned previously, EMA is commonly utilized as a vibration test in a mechanical engineering context. EMA is not governed by a single-degree-of-freedom system, in contrast with the Osstell ISQ<sup>®</sup>, the Periotest<sup>®</sup>, and the Perimeter. EMA—when done correctly—is more versatile, estimating a variety of parameters, including natural frequency, damping ratio, and stiffness. Therefore, it enabled us to determine which parameter exerts the greatest influence on stability.

Different types of bone models are used for purposes of dental implant stability testing *in vitro*. These include animal bones such as porcine bones, human cadaver jaw bones, and polyurethane blocks known as Sawbones<sup>®</sup>, as described in section 2.5 of Chapter 2. Actual human

bones have unique mechanical and physical properties. Using actual human bone in biomechanical testing, however, presents some significant challenges, mainly because of lack of availability but also because of substantial variation between samples. For this reason, animal and synthetic bones are typically used as substitutes. To gauge how effectively Sawbones<sup>®</sup> and porcine bones could be used to accurately represent human mandibular bones, in Chapter 4, we compared the different types of bone models (i.e., porcine bone, human mandibular bone, and Sawbones<sup>®</sup>), using the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> devices, as well as EMA, in addition to measuring apparent bone density. For the samples to be compared on an equal basis, the following were done: First, identical implants were placed into the different bone types, by means of a similar surgical technique, using both dry human and porcine bone samples. Sawbones<sup>®</sup> were then shaped to replicate the geometry of the porcine and human mandibular bone samples. Finally, all samples were tested under similar fixed boundary conditions.

In Chapter 5, we also used micro-CT to estimate bone mineral density (BMD) and the morphometric parameters of both porcine and human bone samples, concluding that porcine bone models probably tend to result in overestimating implant stability. The Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV were unable to differentiate between implants in different types of bone models, suggesting their unreliability for this purpose. Although Sawbones<sup>®</sup> may overall offer a good model for representing human bone, measurements resulting therefrom should be considered with caution, as Sawbones<sup>®</sup> do not accurately reflect actual bone biomechanics. Sawbones<sup>®</sup> are usually selected for achieving comparable biomechanical results, rather than definitive, clinically applicable, information. In addition, the estimated BMD (for human and porcine bone) from micro-computed tomography (micro-CT) fell within the range of the apparent density values.

The morphometric analyses of the human and porcine bone structures were difficult to compare. Because each sample has unique bone patterns, comparing their microstructures was complicated and challenging, as mandibular bone is very different in structure and architecture from femur bone. In addition, the fact that the samples we worked with had implants might have resulted in overestimating the morphometric parameters. Moreover, our samples were derived from old dry bones, which lose a lot of the microarchitecture present in fresh or *in vivo* bones. Additionally, the BIC estimated by the three different methods—BV/TV, i.S/TV, and manual assessment—yielded similar results and was positively correlated to BMD and the natural frequency values, for both types of bone.

In Chapter 6, we examined how experimental setup and boundary conditions affect dental implant stability measurements. Despite the importance of conditions relating to boundary and experimental setup, the literature includes very few studies addressing these two important aspects. Potential issues relating to experimental setup and boundary conditions for *in vitro* tests have for the most part been ignored, as little discussion has taken place in the dental literature on the effects of boundary conditions and experimental setup. Our study of experimental setup involved the use of implants of various lengths, as well as the use of two different types of abutments. Our study of boundary conditions focused on side (lateral) anchorage *in vivo*, as opposed to a bottom-positioned anchorage of the Sawbones<sup>®</sup> blocks using adhesive tape. To calibrate the test model, natural frequency values were measured by means of EMA and then confirmed numerically through use of FEA, for various combinations of both boundary conditions and experimental setup. For purposes of benchmarking through use of the test model, implant stability was estimated using

the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>. To identify the cause of considerable variations observed in the measured natural frequency values, a parametric study was performed through the use of FEA.

We confirmed that boundary conditions play an important role in dental implant stability testing, and should therefore be taken into consideration when interpreting results of such testing. The combined use of EMA and FEA helped us standardize the test models and improve the experimental set up, in three main ways.

**[a] Identification of the residual stresses.** After some trial and error, the first successful version of our test models had an impression coping (IMP) abutment and side-anchorage boundary condition, as seen in Chapter 3. As we replaced the IMP abutment with a locator abutment (LA), we noticed a considerable disagreement between the EMA measurements and the FEA predictions. The measured natural frequency values were also consistently higher than predicted by FEA. After some adjustment, we found that repeated clamping with the vise to retain the side-anchorage boundary condition had created permanent deformation of the Sawbones<sup>®</sup> blocks. The permanent deformation had induced residual stresses, thereby increasing the natural frequency values. This finding implies that the side-anchorage boundary condition is not reliable, as the state of the residual stresses is always unknown. This was the major reason for investigating the bottom-positioned anchorage boundary condition as an alternative.

**[b] Control of critical parameters.** Results of the parametric study indicated the extremely critical nature of the contact area of the bottom-positioned anchorage (i.e., the area between the Sawbones<sup>®</sup> blocks and the metal base), which was created by placing a highly adhesive tape on the metal block, making sure that there were no bubbles, after which we firmly pressed the bottom side of the Sawbones<sup>®</sup> to the tape on the block, ensuring an absence of rocking.



The effect could be on the order of 100-Hz change in the first natural frequency, if the contact area is reduced by 10%.

**[c] Use of repeatability tests.** After controlling the experimental setup for the bottom-positioned anchorage boundary conditions, we found that measurements from the test models with high-density and low-density blocks were fairly consistent with the FEA predictions (e.g., within a 20% margin of error). The measurements from the test models with hybrid blocks, however, did not yield results that were comparable with FEA predictions.

Since all other parameters had been studied and controlled, the most possible cause was variation in Sawbones<sup>®</sup> material. A repeatability test was conducted to determine whether the 10% variations in physical properties of the Sawbones<sup>®</sup> material (disclosed by the manufacturer) had a measurable effect on the natural frequency values. Intact Sawbones<sup>®</sup> blocks with no drilled-in implants of the different standard densities (15-, 40- and 40/20-PCF) taken from various batches were tested. The natural frequency values of these three Sawbones<sup>®</sup> blocks were then measured via EMA and compared for repeatability. This test showed that differences from batch to batch in the hybrid blocks led to sizable variations in natural frequency values.

Using hybrid blocks from a different batch, we then constructed a second test model, using LA and bottom-positioned anchorage. The measured natural frequency values of the second test model were very close to the FEA predictions. If the combined use of EMA and FEA had not been adopted, it would have been very difficult to spot the root cause of the abnormal frequency change. Because they are composite, hybrid blocks may exhibit a larger variation of natural frequency values from batch to batch than the high-density or the low-density blocks.

The hybrid blocks consist of a 2-mm high-density laminate glued on top of a 40-mm low-density substrate. The presence of a glue layer is known to cause natural frequency values to change (Tsay et al., 1999). If the thickness and the stiffness of the glue layer are not controlled precisely from batch to batch, the stiffness and natural frequency values of the entire hybrid block will vary significantly as a result.

The combined use of both EMA and FEA further confirmed that test models with bottom-positioned anchorage yielded more consistent results than those with side-anchorage. Test models with LA yielded higher frequency values than models with IMP abutment. Quantitatively, the EMA results corresponded closely with FEA predictions, with a 20% margin of error.

In addition, through the combined use of EMA and FEA, we proved that the natural frequency values were not affected by changes in dental implant length. If ISQ readings were accurate, they would reflect this observation. In reality, the Osstell ISQ<sup>®</sup> data were somewhat scattered and unable to predict the trend in boundary conditions. The ISQ readings reflected the limitations of using the SmartPeg<sup>®</sup>, making the Osstell ISQ<sup>®</sup> measurements less useful in critical cases. The measured Osstell ISQ<sup>®</sup> values seem to depend heavily on either the cortical layer or the surface that the SmartPeg<sup>®</sup> is closer to (or in contact with). The Osstell ISQ<sup>®</sup> device proved unable to differentiate the trends among the three blocks of different densities.

The findings here relating to the Osstell ISQ<sup>®</sup> are consistent with many critiques previously published in the literature. For example, the Osstell ISQ<sup>®</sup> is purported to have the ability to distinguish between implants under conditions with and without bone loss, an ability unsupported by research findings (Chan et al., 2010). ISQ values have been determined to be not entirely reliable in assessing an integrated implant and the stiffness of the bone-to-implant interface (Nedir

et al., 2004). When osseointegration is in question, ISQ values appear to fluctuate (Ersanli et al., 2005; Friberg et al., 1999; Krafft et al., 2015). It is also difficult to characterize a general standardized range of ISQ readings for a successful implant osseointegration for different implant systems (Ersanli et al., 2005; Krafft et al., 2015) even if the reliability of resonance frequency analysis (RFA) measurement remains low (Krafft et al., 2015; Lachmann et al., 2006). Therefore, the Osstell ISQ<sup>®</sup> cannot be used to accurately provide definitive values for the success, failure, or long-term prognosis of an implant (Haiat et al., 2014).

The Periotest<sup>®</sup> readings for LA were much more consistent and easier to obtain than they were for the IMP abutment, particularly for the hybrid blocks. This could be attributed to the simpler geometry of LA, a solid single piece of metal, 7mm in length. In contrast, IMP abutment is a hollow, two-piece assembly, 12 mm in length, adding more complications to the measurements. As a result, IMP abutment is not as stiff as LA. Therefore, a slight change in the measurement position or angle when using the Periotest<sup>®</sup> could lead to a larger scattering in measured PTV for IMP abutment, compared to LA.

This observation of scattered PTV measurements obtained by the Periotest<sup>®</sup> also agrees with many critiques previously published in the literature. For example, research on the PTV for measuring dental implant stability has reported inability to accurately assess implant stability (Meredith et al., 1998) because of the measure's wide value range (−8 to +50) (van Steenberghe et al., 1995). Moreover, PTV is strongly affected by the rod's direction and position across the length of the abutment (Derhami et al., 1995; Swain et al., 2008). In addition, the limitations and disadvantages of the Periotest<sup>®</sup> instrument include: the impact that clinical variables have on measurements, reduced resolution, low sensitivity, and susceptibility to operator variables when

measuring dental implants (Aparicio et al., 2006; Atsumi et al., 2007; Hobkirk et al., 2006; Kaneko, 1994; Salvi & Lang, 2004). Part of the inadequacy of the Periotest<sup>®</sup> as indicated by various studies could be due to a lack of understanding of how the system being measured responds when activated. It is unclear how the Periotest<sup>®</sup> signals are affected by the interface properties and other physical properties of the measurement system (Aparicio et al., 2006; Atsumi et al., 2007; Hobkirk et al., 2006; Kaneko, 1994; Salvi & Lang, 2004).

From the results presented in this dissertation, we observed that both the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV readings for the low-density blocks were more scattered, indicating that these devices have difficulty measuring implants placed in low-density blocks. Since implant failure is more likely to occur when the bone density is low, it is important that a testing device produce consistent readings—in this case, to indicate impending failure. Therefore, the scattered readings of the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> PTV with regard to low-density blocks are a reflection of their inadequacy for purposes of detecting failing implants.

Many studies have examined the effect of implant length on stability, generating dissimilar results, especially *in vitro*. Although some studies have concluded that implant length has no effect on ISQ values or stability (Gomez-Polo et al., 2016; Meredith et al., 1996; Miyamoto et al., 2005; Monje et al., 2014), others have disagreed, finding either that increasing implant length increases ISQ (Tsolaki et al., 2016), or that increasing implant length decreases stability (Ostman et al., 2006). Our test results indicate that the conflicting conclusions of these studies can be traced mainly to experimental setup conditions. For example, many of these studies (Ahn et al., 2012; Andres-Garcia et al., 2009; Bilhan et al., 2015; Brizuela-Velasco et al., 2015; de Oliveira et al., 2016; Elias et al., 2012; Gomez-Polo et al., 2016; Hong et al., 2012; Lachmann et al., 2006; Mazzo

et al., 2012; Mohlhenrich et al., 2015; O'Sullivan et al., 2000; Romanos et al., 2016; Sennerby et al., 2015; Tabassum et al., 2010; Tsolaki et al., 2016; Wang et al., 2015; Yamaguchi et al., 2015) involved the placement of multiple implants in the same testing block, which we found to cause errors. Most studies used two or three implants of very similar lengths (e.g. 10, 11 and 13mm), (Gomez-Polo et al., 2016; Meredith et al., 1996; Monje et al., 2014) which does not allow for a complete picture regarding comparisons of various implant lengths.

Another reason for conflicting conclusions among previously published studies with regard to the effect of implant length on stability is that measurement of implant stability is generally gauged solely by either the Periotest<sup>®</sup> or the Osstell<sup>®</sup> (Andres-Garcia et al., 2009; Bergkvist et al., 2010; Bilhan et al., 2015; Bornstein et al., 2009; Chan et al., 2010; Chong et al., 2009; de Oliveira et al., 2016; Degidi et al., 2015; Ersanli et al., 2005; Lachmann et al., 2006; van Steenberghe et al., 1995; Winkler et al., 2001). As described above, several problems exist with these devices, pointing to the need for use of other standard tests by which to determine the viability of these devices for purposes of assessing implant stability.

Several clinical studies have shown that implants that are shorter in length tend to have a lower survival rate (Chrcanovic et al., 2014; Winkler et al., 2000). Clinical studies have also shown that implants longer in length last longer (Chrcanovic et al., 2014; Ferrigno et al., 2002). Nevertheless, implant length alone is not a definitive parameter for determining stability, which equates with longevity. The type of bone, position of the implant in the jawbone, and whether the patient is edentulous or dentate—in addition to other factors—collectively play a role in creating the conditions for stability.

To develop and validate a novel and reliable system for estimation of dental implant stability, in Chapter 7, EMA was used to obtain accurate measurements of natural frequency values and linear stiffness of dental implants (with different lengths and widths) in Sawbones<sup>®</sup> models of different densities. FEA was then performed, to model the experimental setup and test environment. The combined use of EMA and FEA facilitated the estimation of angular stiffness coefficient of the dental implant in the Sawbones<sup>®</sup> substrate, serving as an index for quantifying the stability of a dental implant. The results obtained from both the regular platform (RP) and wide platform (WP) implants illustrated that the angular stiffness coefficient is an effective measure to quantify implant stability. This is at the same time easy to understand and intuitive. In practice, when a clinician taps on a dental implant, he or she is getting a sense of the angular stiffness. When a person jiggles his or her own teeth, the person is estimating the stability. The angular stiffness coefficient is theoretically sound because it is almost independent of the size of the abutments. Another advantage of using the angular stiffness coefficient to measure implant stability is that it is not a normalized quantity. This means that we can predict how much the implant will displace under various biting conditions. Moreover, implants of different designs or implants in different bone densities, for example, can be compared and differentiated on an equal basis.

The angular stiffness coefficient, however, is difficult to measure directly in experiments. If natural frequency values or linear stiffness coefficients are measured experimentally, the angular stiffness coefficient may be extracted through use of an accurate mathematical model. More research should be directed into these areas to make measurements or extraction of angular stiffness coefficients more convenient and accurate.

Traditionally, natural frequency values (or resonance frequencies) and length of contact time have been used as measures of implant stability, as with the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup> devices, respectively. As discussed in Chapter 3, natural frequency values and length of contact time do not cover the entire theoretical framework regarding implant stability. In this dissertation, we confirmed that the concept of using natural frequency values to quantify implant stability is inaccurate and obsolete. For example, RP implants of different lengths had roughly the same natural frequency values. Nevertheless, RP implants with a longer implant length had better implant stability because of their higher angular stiffness. Likewise, WP and RP implants had similar natural frequency values but significantly different angular stiffness coefficients.

Since the goal of this dissertation is to quantify implant stability, it is extremely important to secure accurate values of angular stiffness coefficients. This depends on two factors: reliable experimental measurements and an accurate mathematical model. In the studies presented in the preceding chapters, natural frequency values—which have the merits of being both robust and repeatable—were demonstrated to be highly reliable experimental measurements. Linear stiffness coefficients, in contrast, were very sensitive to experimental setup, and the measured linear stiffness values presented some degree of scatter. Therefore, the measured natural frequency values, rather than the measured linear stiffness, present themselves as the best choice for estimating the angular stiffness coefficient.

In addition, the finite element model (FEM) proved to be very accurate, with the model predictions in substantial agreement with the measured natural frequency values. The material properties data from manufacturers were used in the FEM (e.g., density and Young's modulus of

Sawbones<sup>®</sup>) (Table 3.1). These material properties must be used to calibrate the FEM against experimental measurements, in order to prove the model's accuracy.

The measured natural frequency values were proven to be reliable, thereby demonstrating the accuracy of the FEM. We can confidently conclude, therefore, that the angular stiffness coefficients accurately estimated the true angular stiffness coefficients of the experimental setup. This leads to the finding that WP implants have higher stability than RP implants. In addition, stability is positively correlated with dental implant length, whereby implants longer in length exhibited greater stability than those of shorter length.

Linear stiffness coefficient did not prove to be a good measure of implant stability, as this quality was significantly affected by the presence of abutments. Linear stiffness was also very sensitive to any changes in the testing environment. Discrepancies of 50-75% were observed between EMA and FEA predictions when using the high-density (40-PCF) Sawbones<sup>®</sup> blocks for measuring linear stiffness coefficient. Since the linear stiffness coefficient represents a less robust measure, the existence of some disagreement does not mean that the FEM for RP implants in 40-PCF Sawbones<sup>®</sup> was less accurate. In contrast, the FEM was proven accurate because it effectively predicted the natural frequency values. The measurements of the linear stiffness coefficients only imply that the linear stiffness measurements are more reliable for WP and RP implants in 15-PCF Sawbones<sup>®</sup> blocks than for RP implants in 40-PCF Sawbones<sup>®</sup> blocks. Moreover, the agreement of EMA-measured and FEA-predicted linear stiffness coefficients provides further assurance of the accuracy of the finite element model.

This dissertation has endeavored to prove the idea that the angular stiffness coefficient is a viable and reliable method of measuring dental implant stability. All parameters in the experiments



reported here were controlled to the fullest extent possible. The geometry of test blocks, materials used, and boundary conditions were all well controlled. There were some uncontrollable parameters, however, such as the material properties of the Sawbones<sup>®</sup> blocks used in the experiments. Sawbones<sup>®</sup> materials are indeed nonlinear; they have different Young's modulus values in terms of tension and in compression. The force-displacement curves of Sawbones<sup>®</sup> materials are not entirely linear. The homogeneity of Sawbones<sup>®</sup> materials and the uniformity of the Young's modulus values throughout the entire block are also unknowns. These uncontrollable parameters possibly account for the differences between the measured and predicted natural frequency values.

In terms of practical application, the mechanical properties of the bone surrounding the dental implant are unknown. Therefore, the appropriate practice is to adjust the Young's modulus value of the substrate surrounding the implant, so that the natural frequency values predicted by FEA match with the EMA-measured ones, meaning that one needs to fit the stiffness of the substrate materials from the experimental measurements. The fitted substrate stiffness, in combination with the FEM, generates an angular stiffness coefficient that better reflects the experimental setup.

To estimate angular stiffness coefficients, natural frequency values are used to match the EMA measurements and FEA predictions because of their high degree of robustness as experimental measurements. If the linear stiffness coefficients can be measured with equal or better reliability and robustness, one can adjust the Young's modulus value of the substrate surrounding the implant, so that the FEA-predicted linear stiffness coefficients match with the EMA-measured values. Linear stiffness can then be used to estimate angular stiffness.

## 8.2 Highlights of Contributions to The Literature

The published literature on dental implant stability testing is largely comprised of studies characterized by controversy and insufficient information. The studies presented as part of this dissertation represent an attempt to move in the direction of conclusions based on sound scientific findings, taking into account established principles of engineering, so as to enhance dental researchers' understanding about dental implant stability.

A major finding of this dissertation is the value of the combined use of EMA measurements and FEA predictions. Although this is an established methodology in engineering, its use in dentistry is innovative. EMA and FEA results were mutually consistent, both quantitatively and qualitatively. EMA in combination with FEA served as a rigorous and reliable means for calibrating commercial implant stability devices (the Osstell ISQ<sup>®</sup> and the Periotest<sup>®</sup>) and identifying their limitations, as well as helping identify factors necessary for setting up an appropriate test model. EMA and FEA in combination enabled us to not only improve the experimental set-up and standardize the test models but also to create more reliable test models, in addition to assisting with determining the most appropriate boundary conditions.

Most importantly, the combined use of EMA and FEA enabled us to identify a reliable novel method for assessing dental implant stability *in vitro*, based on the demonstrated principle that the angular stiffness coefficient is a highly reliable measure for quantifying dental implant stability. Unlike natural frequency values, the angular stiffness coefficient successfully differentiated between the stability of implants with different lengths and widths, in different

densities of Sawbones<sup>®</sup> models. The angular stiffness coefficient also offers the advantage of being independent of the type of abutment present.

### 8.3 Study Limitations

In this study, the combined use of EMA and FEA made it possible to develop and test a promising novel method for accurately assessing dental implant stability. The angular stiffness coefficient on which the methodology depends, was not measured directly, which would have enhanced the validity of the results.

Our tests were carried out in the context of limited resources, *in vitro*, with implants being placed in synthetic bone material (Sawbones<sup>®</sup>), using a limited number of models. As we have seen in the preceding chapters, however, although Sawbones<sup>®</sup> may overall offer a good model for representing human bone, measurements resulting therefrom should be considered with caution, as Sawbones<sup>®</sup> do not accurately reflect actual bone biomechanics. Sawbones<sup>®</sup> are usually selected for their ability to achieve biomechanical results comparable to that of real bone, rather than for definitive, clinically applicable, information. In addition, with *in vitro* testing, parameters and boundary conditions can be controlled. With *in vivo* testing, it is sometimes hard to control parameters, as they may function differently. Although an *in vitro* test is efficient for testing scientific theories, *in vivo* tests are needed for purposes of producing more clinically pertinent outcomes. These *in vitro* test models enabled us to estimate primary stability only, as secondary stability would have required osseointegration by means of *in vivo* experiments.

For the studies presented in this dissertation, we were limited to using one particular implant system (the Branemark MKIII), with two different widths and various lengths, in addition

to two different abutments (impression coping [IMP] and locator abutments [LA]). Branemark dental implants were used in this study primarily because the standard Branemark screw implant was the first implant design on the market and has been used for more than 45 years. According to the manufacturer, Nobel Biocare, Branemark is the most scientifically documented implant system. These two abutments—IMP and LA—were used to study the effect of abutment length on dental implant stability. However, numerous abutments as well as a wide variety of implant designs exist. Implants with different thread, body shape, and thread design (e.g., thread geometry, thread pitch, thread depth/height, thickness/width or thread helix angle) are available. In addition, implants vary in terms of materials, surface morphologies, and surface coatings (Abuhussein et al., 2010; Geng et al., 2004a; Geng et al., 2004b). Other current endosseous implant systems come in a tapered shape or are available as solid or hollow screws or cylinders. To validate our findings, they should be tested on the complete range of implant systems available.

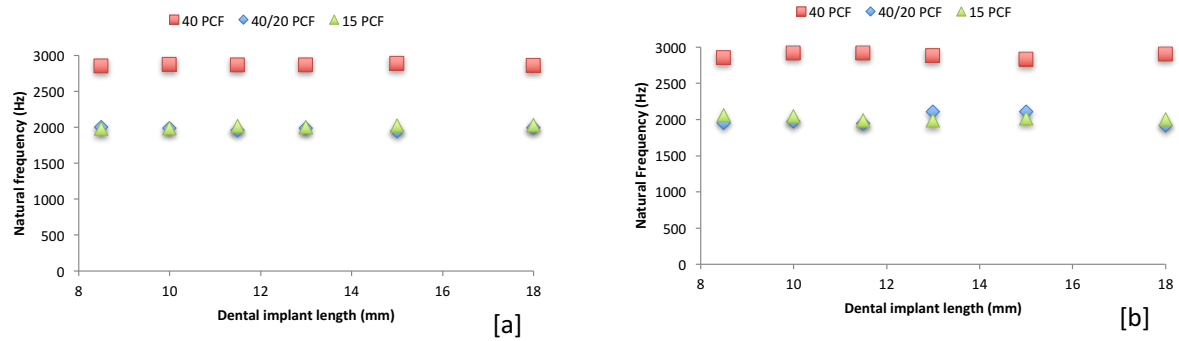
This dissertation focused on a select number of biomechanical factors affecting implant stability, including varying the implant length and diameter, abutment length, and bone density. However, other factors representing more clinical situations, such as effect of the presence of soft tissue at the bone-to-implant interface and bone loss around the implant, would need to be further investigated using angular stiffness, to fully validate our findings. Although we did examine implants placed in different bone types in Chapters 4 and 5, we did not estimate their stability. In addition, we estimated morphometric parameters in *in vitro* non-integrated dental implants and bone, which may differ significantly from *in vivo* osseointegrated implants.

## 8.4 Suggestions for Future Research

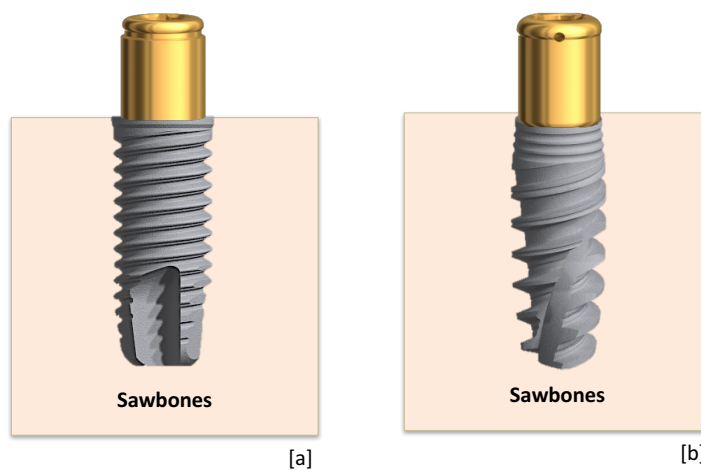
The findings of the studies described in the preceding chapters present a number of ideas for future research. Some of the most promising are outlined below.

### 8.4.1 *Estimate the angular stiffness of different implant designs.*

In this dissertation, the screw-type implants of one particular system (Branemark MKIII) were used to test our hypotheses. Future research might include estimating the angular stiffness for a variety of other implant systems. Preliminary results have shown that natural frequency values estimated by means of EMA did not allow for the differentiation of differences between the Branemark MKIII implants and the Nobel Active implants, as illustrated by Figure 8.1. These two implant designs differ significantly in the thread design, with Nobel Active having more aggressive (deeper) threading. Moreover, Nobel Active implants are more tapered toward the apex, while the Branemark implants have parallel walls and three cutting chambers, as shown in Figure 8.2. Future studies could involve creation of FEMs for these two different implant types, for which to estimate the angular stiffness coefficient of the implants placed in these models, based on the natural frequency values obtained by EMA.



**Figure 8.1** EMA natural frequency values were very similar for two different implant models in three different Sawbones densities: [a] Branemark and [b] Nobel Active.

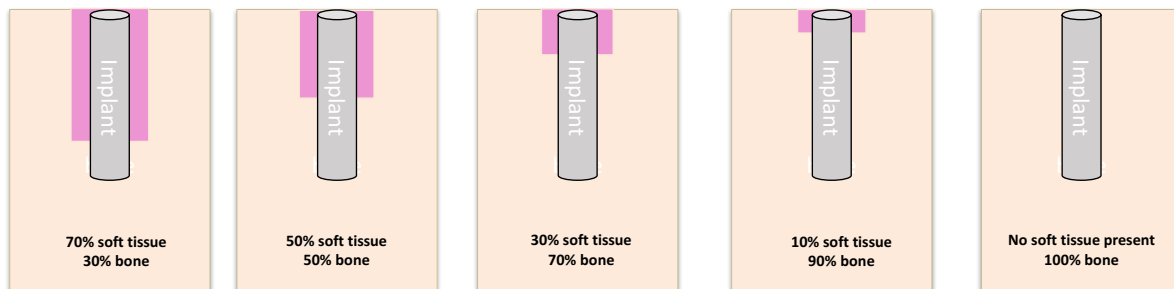


**Figure 8.2** Comparison of two implant designs: [a] Branemark and [b] Nobel Active.

8.4.2 *Study the different biomechanical factors affecting implant stability, using angular stiffness coefficient.*

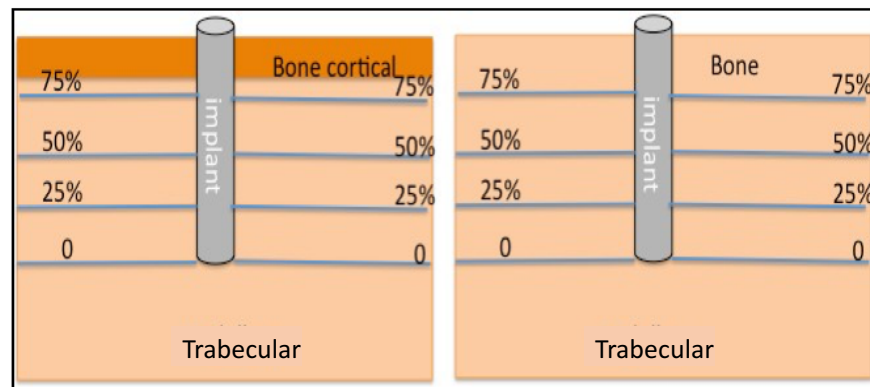
Investigation of factors such as the effect of the presence of soft tissue at the bone-to-implant interface and bone loss around the implant, by using the angular stiffness coefficient, would enhance our understanding of implant stability.

To evaluate the effect of the presence of soft tissue on angular stiffness, the amount of soft tissue material surrounding the implant could be varied incrementally (30%, 50%, 70%, 90%, and 100%) in a Sawbones® model (with 30% indicating 70% soft tissue and 100% meaning no soft tissue). Silicon material could be placed between the implant and Sawbones®, as illustrated in Figure 8.3.



**Figure 8.3** An example of test models that could be used to estimate the effect of the presence of soft tissue between the implant and Sawbones on the angular stiffness coefficient, with the percentage of soft tissue varied incrementally (30, 50, 70, 90, 100).

To evaluate the effect of bone loss around the implant, the percentage of bone support around the implant could be varied incrementally (100%, 75%, 50%, and 25%), as illustrated in Figure 8.4.



**Figure 8.4** An example of test models that could be used to evaluate the effect of bone loss around the implant, the percentage of bone support around the implant could be varied incrementally (of 100%, 75%, 50%, and 25%).

Although we did examine implants placed in different bone types, as reported in Chapters 4 and 5, we did not estimate their stability. The next step would call for creating FEMs for these bone samples. To create accurate FEMs, the Young's modulus values of these different bone models should be estimated through compression and tensile tests. After that, the angular stiffness coefficients of the implants placed in these bone models could be estimated, based on the natural frequency values obtained by EMA.

Various implant designs could be tested—and different bone densities could be experimented with—in the process of carrying out the studies outlined above.



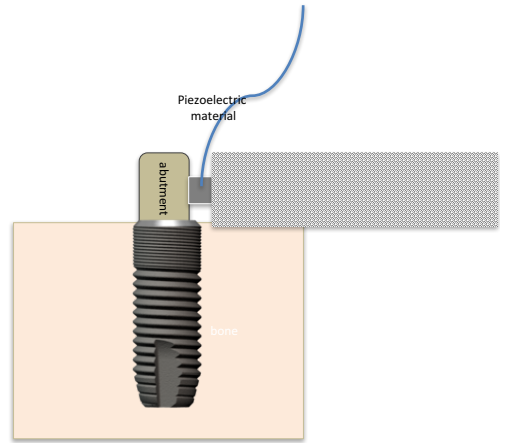
#### 8.4.3 *Experiment with a prototype (patent pending) for estimating dental implant stability, using the angular stiffness coefficient.*

The patent pending notation refers to the development of a device that would estimate the angular stiffness coefficient from measured natural frequency values or linear stiffness values of dental implants placed in bone models or in clinical settings, so as to determine their stability, would potentially be of substantial benefit to clinicians. One example of how this may be accomplished on the part of the pending patent holders would be the use of electromechanical models such as those made from piezoelectric material. Piezoelectric refers to the ability of certain materials to generate an electric charge in response to applied mechanical stress. A piezoelectric ceramic block could be attached to the tip of a rigid rod, with the block held against the implant abutment and a voltage applied to the block, causing both the implant and the block to deform, as the electrical current through the block changes. Measuring the electrical impedance through the piezo-block would allow for natural frequency values and linear stiffness coefficient to be estimated (Figure 8.5 [a]).

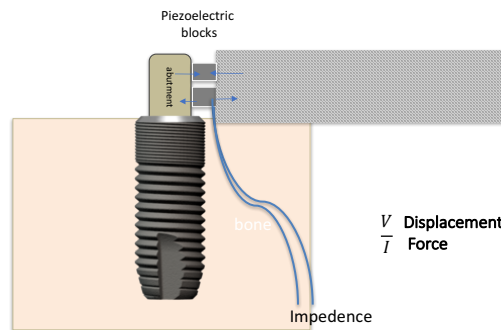
Another example of the possible use of electromechanical models (EM) on the part of the pending patent holders would be the inclusion of a pair of transducers (e.g., piezo-blocks) at the tip of a rigid rod, with the two blocks driven in an out-of-phase manner, to generate angular excitation, causing the difference of the measured impedances to be proportional to the angular stiffness coefficient of the implant (Figure 8.5 [b]). In addition, the previously described devices could be used in combination with a FEM of an implant, which could be a database of finite element simulations of various commercially available implants. The computer model would convert the stiffness measured by the EM device to stability of the dental implant (e.g., % of soft

tissue surrounding the implant). Such a prototype could further be used in *in vitro* studies, to estimate angular stiffness coefficient, as suggested above, and compared to the benchmark (EMA and FEA).

When perfected, this prototype could be used by others to estimate stability of various implant designs in animal models (e.g., canine and porcine), allowing secondary stability to be estimated. The estimation of angular stiffness coefficient could be compared to histomorphometric analysis of the bone-to-implant interface or morphometric analysis, estimated by means of microcomputed tomography (micro-CT). This prototype could be used to estimate angular stiffness in clinical settings, for dental implant patients, at various stages of healing. The effect of different designs, bone type, and location of the implant in the jaw bones— in combination with the age and various health characteristics of the patient—could all be parameters that might affect the stability of the implant.



[a]



[b]

**Figure 8.5** Examples of prototypes (patent pending) that could be used to estimate angular stiffness. [a] A piezoelectric ceramic block could be attached to the tip of a rigid rod, with the block held against the implant abutment and a voltage applied to the block, causing both the implant and the block to deform, as the electrical current through the block changes. Measuring the electrical impedance through the piezo-block would allow for natural frequency values and linear stiffness to be estimated. [b] A pair of transducers (e.g., piezo-blocks) at the tip of a rigid rod, with the two blocks driven in an out-of-phase manner, to generate angular excitation, causing the difference of the measured impedances to be proportional to the angular stiffness of the implant.

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## Education

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- 2017      **University of Washington** — Seattle, WA, USA  
Doctor of Philosophy (Ph.D.) Oral Health Sciences  
Thesis title: In Search of An Improved Methodology to Measure Dental Implant Stability: Combing Experimental Modal Analysis and Finite Element Analysis to Reach Maximum Reliability.  
Supervised by: John Sorensen  
GPA: 3.85/4
- 2010      **Indiana University-Purdue University** — Indianapolis, IN, USA  
Master of Dental Science (MSD) Dental Materials  
Thesis title: Bone Regeneration Around Novel Titanium Implants.  
Supervised by: Tein-Min (Gabriel) Chu  
GPA: 3.6/4
- 2005      **King AbdulAziz University** Jeddah, Saudi Arabia  
Bachelor of Dental Medicine and Surgery (BDS)  
Grade: (B) (Very Good with Second Honors Degree).  
GPA: 80.6/100 4.2/5

## Work Experience

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- 05/2007 - Current      **King AbdulAziz University, Faculty of Dentistry** — Jeddah, Saudi Arabia  
**Junior Instructor/ Teaching Assistant in Dental Biomaterials**
- 01/2007 - 03/2007      **King Fahad General Hospital** — Jeddah , Saudi Arabia  
**Dental Trainee**
- 09/2005 - 08/2006      **King Abdul Aziz University, Faculty of Dentistry** — Jeddah, Saudi Arabia  
**Dental Intern**
- 03/2006 - 05/2006      **King AbdulAziz University, Faculty of Dentistry** — Jeddah, Saudi Arabia  
**Assisting in the supervision of the 4th year dental students during there Operative Dental Clinics**
- 07/2004      **Saudi German Hospital** — Jeddah , Saudi Arabia  
**Summer Trainee**

## Research Experience

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- **December 2012 - April 2017 In search of an improved methodology to measure dental implant stability: combining experimental modal analysis and finite element analysis to reach maximum reliability.**  
I.Y. (Steve) Shen's lab, Department of Mechanical Engineering, University of Washington, Seattle, WA, USA.  
John A Sorensen's lab Biomimetics, Biomaterials, Biophotonics, Biomechanics and Technology, Department of Restorative Dentistry, School of Dentistry, Seattle, WA, USA.
- **June 2011 - October 2012 Use of non toxic shape memory alloys as dental implants.**  
Minoru Taya's lab. Department of Mechanical Engineering, University of Washington, Seattle, WA, USA.
- **April - June 2012 Effect of Au addition on shape memory behavior of Ti-Au-Cr-Zr biomedical alloys.**  
Hideki Hosoda's and Tomonari Inamura's lab. Precision and Intelligence Laboratory, Tokyo Institute of Technology, Yokohama, Japan.
- **March - June 2011 Methacrylated hyaluronan and poly(vinyl alcohol) bi-layered scaffolds for periodontal tissue regeneration. (Lab. rotation project).**  
James D. Bryers's lab. Department of Bioengineering, University of Washington, Seattle, WA, USA.
- **September 2010 - March 2011 Metal-Titanate complexes as novel antibacterial compounds for oral-facial infections. (Lab rotation project).**  
Sun O Chung's lab. Department of Oral health sciences, School of Dentistry, University of Washington, Seattle, WA, USA.

- **July 2009- June 2010 Bone regeneration in novel dental implants (MSD Project).**  
 Tein-Min (Gabriel) Chu's lab. Department of Dental materials, School of Dentistry, Indiana University-Purdue University Indianapolis, Indianapolis, IN, USA.
- **July 2009- February 2010 *In vivo* evaluation of novel dental implants from rapid prototyping.**  
 Tein-Min Gabriel Chu's lab. Department of Dental Materials, School of Dentistry, Indiana University-Purdue University Indianapolis, Indianapolis, IN, USA.
- **July 2007-May 2008 Repair and replacement perception of dental restorations done.**  
 With Mohammed K Yousef, Department of Conservative Dentistry, Faculty of Dentistry, King Abdulaziz University, Jeddah, Saudi Arabia.
- **March 2006- March 2007 Prevalence of congenital dentofacial defects among children of Jeddah and Makkah cities.**  
 Fahd Ba Nasr and Al-Sayed Nawar, Department of Prosthodontics, Faculty of Dentistry, King Abdulaziz University, Jeddah, Saudi Arabia. (Participated in data collection).
- **September 2004-April 2005 Prosthodontic needs of the aging patients in geriatric homes of Jeddah and Makkah cities.**  
 Fahd Ba Nasr and Al-Sayed Nawar, Department of Prosthodontics, Faculty of Dentistry, King Abdulaziz University, Jeddah, Saudi Arabia. (Participated in data collection).
- **September 2003- March 2004 Effects of fasting on glucose-regulating hormones.**  
 Hamed I Khouja, Faculty of Allied Sciences, King AbdulAziz University, Jeddah, Saudi Arabia. (Participated in volunteer's recruitment and data collection)

## Publications

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**Khouja, N.**, Tai, W., IY, S. & Sorensen, J. (2017a) Dental Implant Stability *In Vitro*: A Novel Method to Quantify Implant Stability. J biomechanics (In Preperation).

**Khouja, N.**, Tai, W., Shen, I. & Sorensen, J. (2017b) A Critical Analysis Of Dental Implant Stability Measurement: Biomechanical Principles, Experimental Modeling and Experimental Measurement Data. (In preparation).

**Khouja, N.**, Tai, W., Shen, I. & Sorensen, J. (2017c) Dental Implant Stability: Effect of Boundary Conditions and Experimental Setup. (In preparation).

Chu, T.G., **Khouja, N.**, Chahine, G., Kovacevic, R., Koike, M., et al., (2016) In vivo Evaluation of a Novel Custom-Made Press-Fit Dental Implant Through Electron Beam Melting® (EBM®). Int J Dentistry Oral Sci. 3(11), 358-365.

Yousef M, **Khoja N.** (2009) Repair and Replacement Perception of Dental Restorations. Journal of King AbdulAziz University: Medical Science 16(2), 75-85.

## Presentations

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**Khouja N**, Sorensen J. Effect of Implant Length and Abutment Material on Implant Stability. Presented at the AADR Annual Meeting & Exhibition. March 19-22, 2014, Charlotte, NC.

**Khouja N**, Shinohara Y, Tahara M, Inamura T, Hosoda H, Taya M. Effect of Au Addition on Ti-Au-Cr-Zr Biomedical Alloy's Properties. Presented at IADR/AADR/CADR General Session & Exhibition. March 20-23, 2013, Seattle, WA.

**Khouja N**, Chu TM G, Chahine G, Kovacevic R, Koike M, Okabe T. In vivo Evaluation of Novel Custom-Made Press-Fit Implants. Presented at the IADR General Session. July 14-17, 2010, Barcelona, Spain.

**Khouja N**, Chu TM G, Chahine G, Kovacevic R, Koike M, Okabe T. In vivo Evaluation of Novel Experimental Implants. Presented at the AADR Annual Meeting. March 3-6, 2010, Washington DC.

**Khouja N**, Yousef M. Prevalence of replacement and repair of dental restorations. Presented at the 2nd International King AbdulAziz University and the 19th Saudi Dental Society Conference. March 2008 Jeddah, Saudi Arabia.

## Patents

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U.S. Provisional Patent Application No. 62/469,854 filed 3/10/2017  
Entitled: "Methods to Measure and Evaluate Stability of Dental Implants and Teeth"  
Inventors: I-Yeu Shen, John Sorensen, Naseeba Khouja, Wei Che Tai  
UW Reference: 47789.01US1

## Awards

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July 2010 AADR Travel Block Award.  
2006 Academic Distinction Award, King AbdulAziz University, Jeddah, Saudi Arabia.  
2002 Academic Distinction Award, King AbdulAziz University, Jeddah, Saudi Arabia.  
2001 Academic Distinction Award, King AbdulAziz University, Jeddah, Saudi Arabia.