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ORIGINAL ARTICLE

Characterisation of bone following ultrasonic cutting

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Abstract
Objectives Ultrasonic surgery is an increasingly popular technique for cutting bone, but little research has investigated how the ultrasonic tip oscillations may affect the cuts they produce in bone. The aim of this investigation was to evaluate the oscillation and cutting characteristics of an ultrasonic surgical device.
Materials and methods A Piezosurgery 3 (Mectron, Carasco, Italy) ultrasonic cutting system was utilised with an OP3 style tip. The system was operated with the tip in contact with porcine bone samples (loads of 50 to 200 g) mounted at 45° to the vertical insert tip and with a water flow of 57 ml/min. Tip oscillation amplitude was determined using scanning laser vibrometry. Bone surfaces defects were characterised using laser profilometry and scanning electron microscopy.
Results A positive relationship was observed between the magnitude of tip oscillations and the dimensions of defects cut into the bone surface. Overloading the tip led to a reduction in oscillation and hence in the defect produced. A contact load of 150 g provided the greatest depth of cut. Defects produced in the bone came from two clear phases of cutting.
Conclusions The structure of the bone was found to be an important factor in the cut characteristics following piezosurgery.
Clinical relevance Cutting of bone with ultrasonics is influenced by the load applied and the setting used. Care must be used to prevent the tip from sliding over the bone at low loadings.

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Introduction

Advances in ultrasonic technology have led to the introduction of specialised surgical systems which are used for dental implant site preparation, osteoplasty procedures (surgical repair or alteration of bone), bone chip harvesting and hand and sinus surgery [1, 2]. Piezoelectric technology is the preferred generation of ultrasound for bone cutting purposes as it allows the use of light and easy to handle handpieces with sterile water-cooling systems. Such instruments have been adapted so that they may reach previously inaccessible or hard to reach areas of the bony skeleton. Clinical observations have shown that ultrasonic instruments provide precision cutting of the bone and require less operating pressure from the clinician [3, 4]. Biophysical phenomena produced by the ultrasonic tip motion such as cavitation and acoustic microstreaming may also aid the bone cutting in a similar manner to ultrasonic scaling [5, 6]. Other clinical findings include a smoother finish, faster healing of bone and less facial swelling following the use of an ultrasonic surgical procedure [3, 4].

Many articles are clinically focussed and report on the surgical qualities of the instrument providing useful qualitative information on how the instruments perform. However, tip movement depends strongly on its geometry, and work on ultrasonic scalers has shown that it is the design that influences function [5]. Commercially available instruments are designed to reproduce conventional bone cutting designs such as chisels or rotary devices. A recent study looked at the performance of a Piezosurgery bone cutting tip under a range of operating conditions and attempted to correlate the vibration patterns of the tip with the defects produced on a bovine bone sample [7]. It was shown that nature of the surface (i.e. cancellous or cortical bone) being cut might significantly alter the mode shape and magnitude of the probe oscillation. The hypothesis of this investigation is that the oscillation patterns of a universal surgical tip design will have a direct effect on the type of cutting that is produced whilst in contact with the bone.

Materials and methods

A Piezosurgery 3 (Mectron, Carasco, Italy) ultrasonic surgical system was used with a single OP3 insert in both the 'cortical' and 'spongious' surgical programme power settings. The OP3 tip was selected as a universal design as it is used for periodontal osteotomy (surgery on the supporting tissue surrounding the teeth), bone chip harvesting and inflammatory tissue removal. The tip is spoon-shaped and constructed from hardened stainless steel. The handpiece was positioned and secured with the anterior surface of the insert presented perpendicularly to the camera of a 1D scanning laser vibrometer (SLV; PSV-300-F/S High Frequency Scanning Vibrometer System, Polytec GmbH, Waldbronn, Germany). The laser of the SLV was focused onto the tip to ensure a good signal-to-noise ratio. The scanning area on the tip was defined using a fully adjustable virtual measurement grid and comprised of a single column of 40 scan points over the anterior face of the tip.

The SLV software can display the oscillatory motion of the tip as a colour-coded map superimposed on a captured video image of the target tip. Vibration displacement amplitude and frequency data and spatial location of each point scanned are also presented by the software. A frequency measurement range of 0–100 kHz was selected to allow detection of the fundamental operating frequency of the tip (24–29 kHz) and any higher order harmonics using a swept-sine waveform. The system was set to perform 800 FFT calculations, giving a frequency resolution of 125 Hz, thus enabling the fundamental oscillation frequency of the tip to be accurately determined.

Prior to obtaining measurements, the laser was focused and aligned on the anterior surface of the OP3 tip. A scanning area was then defined on the tip using the fully adjustable virtual measurement grid feature in the SLV software. A single column consisting of 25 individual scan points was defined from the free end of the tip to the top of the anterior face.

All measurements were taken in phase with each other, and a reference signal was used to achieve this as described previously [8]. Briefly, the reference signal for the ultrasonic instruments consisted of an iron core with a coil placed adjacent to the handpiece. Through mutual induction, the change of current in the piezoelectric handpiece induces an e.m.f. in the coil which is proportional to the change of current. Each time the maximum amplitudes are generated, the SLV is triggered to perform a measurement at the set coordinates [8]

Unloaded and loaded measurements

Scans were taken with the tip unloaded oscillating freely in air with water flowing over the tip at a rate of 57 ml/min. Five repeat scans were taken for both the cortical and spongious programme settings. Each scan lasted approximately 5 s with a 5-s interval between scans.

For loaded measurements, the insert tip contacted a freshly obtained porcine bone sample taken from the femur. The bone was cut using a bone saw to provide both a cortical and spongious sample. The bone was placed in a holder (Fig. 1), which allowed a bone sample of approximately 3×3 cm to be oriented at angles from 0° to 90°. The handpiece was clamped vertically and prior to the tip contacting the bone; the insert was operated in free air for 5 s to allow it to reach operating speed. Initially, a 50-g load of tip to bone was maintained using a pan balance.

The OP3 insert was positioned vertically to facilitate scanning, and the bone sample was held using the specially designed bone holder at a 40° angle (Fig. 1). Scanning took

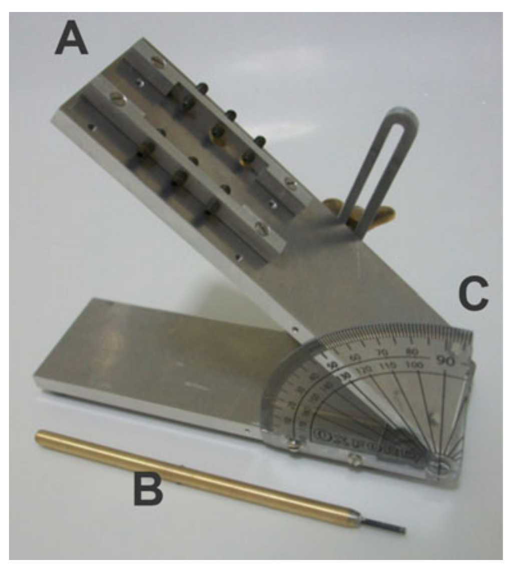

Fig. 1 Bone holding device for loaded measurements. The bone sample is held in the holder (**a**) whilst the angle of the bone to the drill is adjusted via screwdriver (**b**) and referenced to a protractor (**c**)

place along the longitudinal axis of the tip. For both the cortical and spongious bone samples, five repeat readings were taken for 50- to 200-g loads (5 s duration) with the water flow rate over the working tip set at 57 ml/min. Student's t test was used to determine differences between the displacement amplitude oscillations of the two tip designs.

Laser profilometry

The bone sample defects produced by the OP3 tip were scanned using a TaiCaan Xyris 4000 WL/CL 3D metrology system (Taicaan Technologies Ltd, Southampton, UK). The system was operated in the 'confocal laser' mode to allow a noninvasive evaluation of the surface topography. The system consists of a precision laser (10 nm resolution) oriented vertically above a movable stage upon which the sample may be moved across the laser beam. The laser and stage structure is mounted on a granite support to provide vibration damping. Scan parameters were set to provide a high-resolution scan of the entire bone sample using 401×401 point scanning grid. Each scan lasted approximately 24 h.

Analysis of the scans was performed using the Taicaan Boddies software (version 2.05) general-purpose analysis tool. Using this software, any underlying curvature of the bone sample surface was removed by applying a second-order polynomial to the surface. Using the Boddies 2D graphical display, profiles across each cut were taken using the 'free' line profiling option, and the width, length and volume of each cut were evaluated. Profilometry analysis was conducted for the cortical and spongious bone samples. The spongious sample was gold coated (30 nm thickness) to increase the reflectivity of the sample (K550X Sputter Coater, Quorum Technologies Ltd, Kent, UK) and assist laser profilometry to overcome variations in the topography and trabecular structure.

Images of the cortical and spongious samples were obtained using a scanning laser microscopy scanning electron microscope (SEM) with an incident beam voltage of 20 kV with ×50 magnification for the cortical setting and at 20 kV with ×75 magnification for the spongious setting.

Results

Oscillation of the tip

A typical oscillation mode shape for the unloaded tip, plotting mean displacement amplitude along the length of the anterior surface of the insert for cortical and spongious modes, is shown in Fig. 2. The maximum displacement amplitude does not occur at the free end of the OP3 tip and was observed at the 7- to 9-mm position. At high loads, it was not possible to fully observe the edge of the rounded tip with the SLV, and this led to the apparent decrease for those readings in this area. Nodal points, where the oscillation of the tip is a minimum, were observed in both modes of operation, approximately 3 mm from the free end of the tip (Fig. 2).

The OP3 tip was contacted against cortical and spongious porcine bone samples using loads of 50 to 200 g. Oscillation mode shapes along the length of the OP3 tip were produced for both the cortical and spongious mode of operation (Fig. 2) with maximum displacement amplitude often occurring at the 8–12-mm position.

The displacement amplitude (half peak to peak amplitude) for each operating mode is shown in Table 1. The displacement amplitude of the tip decreases as it contacts cortical bone under increasing load. There is no overall change in the oscillation when the tip contacts the spongious bone under increasing load. It is only at the highest loadings of 200 g that the tip oscillation is significantly reduced when contacting cortical bone in comparison to spongious bone (p=0.003)

Bone defect laser profilometry

The defects produced by the OP3 tip on the cortical and spongious bone under the four contact loads were analysed using laser profilometry. Typical scans of the surface characteristics are colour-coded according to surface height (Fig. 3). Typical sections through the bone cuts are shown in Fig. 4 providing a 2D view of the defect dimensions. There was an initial contact phase as the tip slides across the surface and then a secondary phase where the deepest cut is produced (Fig. 4a). The width of the deepest defect produced was used for measurement purposes. The length of the defect produced (Fig. 4b) followed a saucer shape mirroring the design of the metal tip. Using these images, the average defect depth, width, length and volume of the cortical defects were obtained (Table 1). It was not possible to measure the depth of the spongious surface, and these measurements were limited to defect width and length.

When the tip contacts cortical bone, then it quickly produces an increase in depth from loads of 50 to 100, but there is a decrease at 200 g, which mirrors the decrease in displacement amplitude. Qualitatively, the cutting of cortical bone produces a narrow width compared to the spongious bone.

SEM images of the defects produced under the increasing contact loads are shown in Fig. 5. The defect may be seen to increase in size, and the creation of an initial contact phase of the cut is outlined. Whilst it is possible to clearly see the defect on the outer cortical bone of the samples, the spongious bone structure did not allow such a clear demarcation of the cut. The comparison of the cortical and spongious bone cuts is shown in Fig. 6.

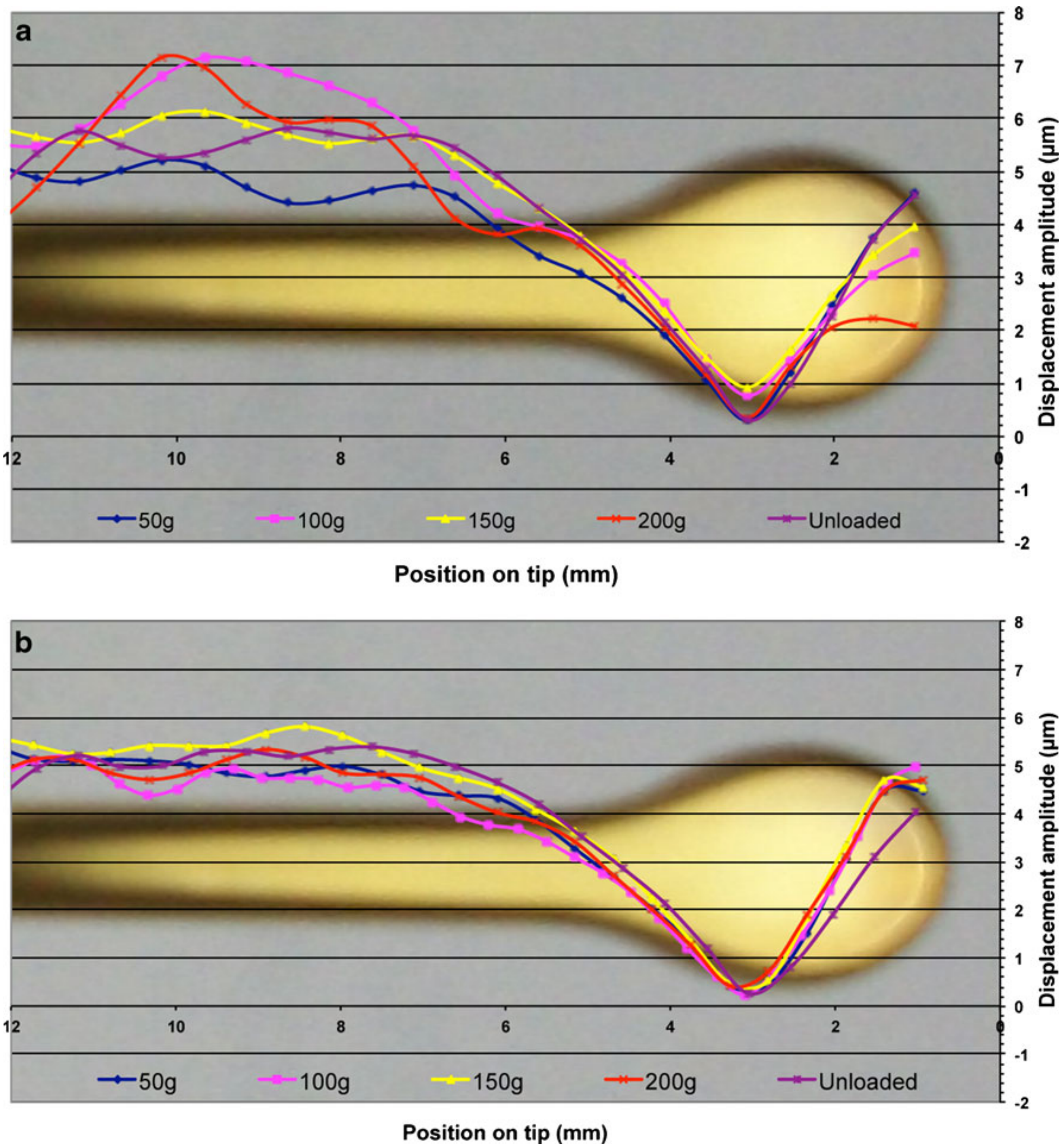

Fig. 2 Mean displacement amplitude along the length of the OP3 insert for the cortical (**a**) and spongious (**b**) modes superimposed over a scaled image of an OP3 insert

Discussion

Ultrasonic cutting of hard tissues has become established as a preferred surgical tool especially where small precision cuts are required and in difficult and previously inaccessible areas. The choice of different tip designs is very attractive to surgeons. The tip used in this study is commonly used in dental implant surgery for harvesting bone or preparing bone sites for the placement of implants. However, other surgical specialities such as otology and orthopaedics have now adapted the use of such cutting tools. Although the literature contains many successful case studies using such equipment, there is a need to scientifically evaluate how these devices work and investigate whether their performance can be optimised in relation to cutting of the bone. Previous work using a direct ultrasonic cutting tool, with a saw-toothed edge, demonstrated that the cancellous bone tended to damp down the oscillations [7]. The drill produced cuts in the bone, and measurement of the resulting defects showed that the tip cut into the bone up to depths of 0.36 mm. The optimal cutting for this particular design of tip was when the tip was in contact with the bone with a load

Table 1 Mean maximum displacement amplitudes (±1 standard deviation) at the free end of the OP3 insert (first antinode) for both operating modes of the generator

Operating mode	Load (g)	Mean displacement amplitude (μm)	Defect depth (mm)	Width (mm)	Length (mm)	Volume (mm^3)
Cortical	50	4.60±0.47	0.103±0.027	0.141±0.070	1.210±0.004	0.004±0.002
	100	3.47±1.14	0.184±0.022	0.176±0.025	1.765±0.105	0.011±0.004
	150	3.97±1.12	0.212±0.024	0.229±0.021	2.046±0.036	0.017±0.004
	200	2.85±0.74	0.190±0.017	0.192±0.036	2.089±0.048	0.020±0.003
Spongious	50	4.47±1.06		0.780±0.109	2.011±0.099	
	100	4.99±1.51		0.800±0.097	2.042±0.403	
	150	4.55±0.88		1.103±0.118	2.385±0.699	
	200	4.71±0.86		1.013±0.103	2.709±0.535	

Bone defect depths, widths and lengths are provided for the cortical setting. For the spongious settings, depth data could not be obtained; thus, only defect length and width are provided

of 100 g [7]. In comparison, the rounded spoon-shaped tip used in this study produced cuts of 0.18 mm in the cortical bone and has lower displacement amplitudes for the same settings (Table 1). The rounded tip design cuts into the bone to retrieve "shavings" for harvesting. The intial contact phase seen in the SEM views (Figs. 5 and 6) shows this cutting action prior to the tip moving down into the bone to produce a deeper cut. This was different to the previous study where application of load damped the oscillation amplitude of the tip in the cortical mode but not in the spongious mode. In comparison, the tip in the present study had a solid edge and not a saw-toothed edge. A saw-toothed edge tip is not able to free itself quickly when contacting trabecular bone, whilst this did not prevent the solid shaped tip from oscillating and cutting into the bone.

The observation of maximum displacement amplitude occurring in the 7- to 9-mm position with the tip unloaded

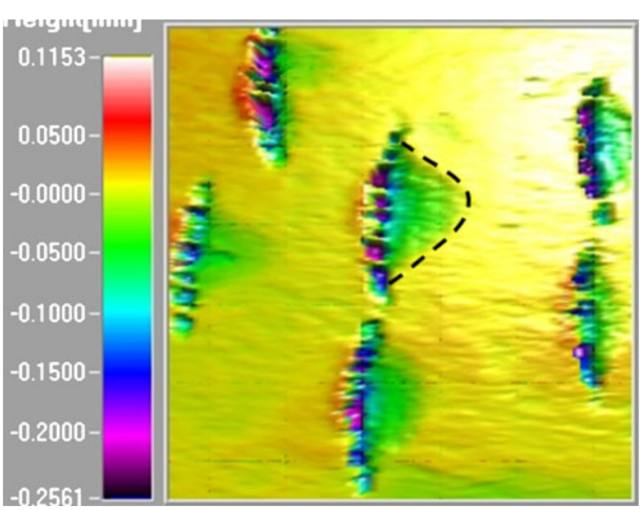

Fig. 3 Surface profiles of the cortical and spongious bone surfaces obtained using laser profilometry. Depth is represented by scaling according to colour as shown

indicates that the design of the tip warrants further investigation. As the load increases, the energy supplied to drive the tip is distributed further up the tip, thus explaining why larger oscillations are seen further away from the free end of the tip when under load. This change in the oscillation of the tip merits further investigation.

For the cortical bone cutting setting, the defect depth relates to the oscillation amplitude of the tip at loads of 100 g and greater. At 50 g, it is likely that the tip is not contacting against the bone for optimal cutting to take place, and the tip slides over the surface before catching the bone and then producing a cut. Therefore, a design that prevents this sliding would be more efficient and merits further investigation. The cut profiles in Fig. 4 highlight that there are two phases of cutting—an initial shallow phase as the tip slides across the surface of the bone and then a second, deeper, phase as the tip 'bites' and cuts vertically.

The defect width is related to the oscillation amplitude of the tip at loads of 100 g and greater. The defect depth increases and decreases with the displacement amplitude (Table 1). At 200 g, the defect length still increases. Whilst the longitudinal motion of the tip is reduced by damping, there is space for the lateral component of the tip motion to increase, thereby increasing the length of the defect. For the spongious bone cutting setting, the defect length is increased with the oscillation amplitude of the tip at all loads. This may be due to the trabecular structure of the bone which frees up the motion of the tip.

Measurements of the spongious surface were limited to defect width and length due to poor definition obtained during the laser profilometry measurements, but a similar pattern to the cortical bone was seen (Table 1). The SEM images of the cuts produced in cortical bone for contact loads of 50 to 200 g demonstrate the distinctive initial cut (Fig. 5) with the secondary deeper cut. The differences in the cuts produced between cortical and spongious bone are

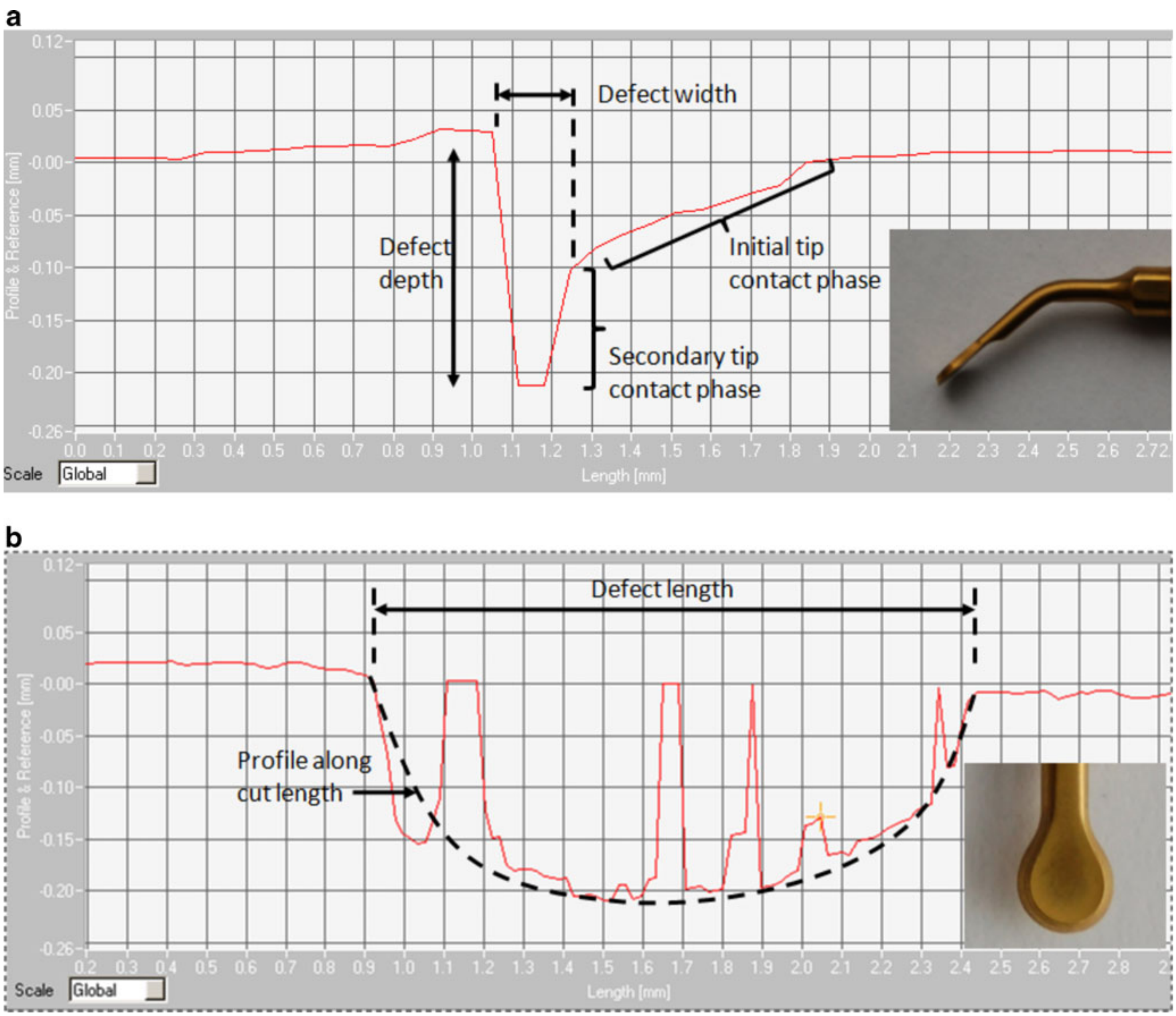

Fig. 4 Defect width (**a**) and defect length (**b**) profiles obtained from the profilometer. The peaks shown in the image (**b**) are artefacts produced by the profilometer software

highlighted in Fig. 6, which shows an SEM of a cut on the spongious surface. Whilst the cut is clearly defined in the cortical bone, there is less definition possible for the spongious bone, and the profilometer was only able to measure the width and length but not the depth and volume of the cuts.

The ultrasonic cutting tool has been received enthusiastically in the literature, and there have been several clinical reports on its use [1, 2, 4, 9]. However, most of these reports are based on clinical observations, and the physical mechanisms are not reported. There have been observations on the potential damaging effect in vitro that the use of ultrasound may have on the bone. A study on pig cadavers showed that the cut surface of bone following ultrasonic instrumentation was not well defined, and this differed to other techniques such as rotary or laser instrumentation [10]. Our SEM results were focused on ultrasonic cutting only and did not compare the efficiency to other devices such as rotary drills and merits further investigation. In order to make the ultrasonic inserts more effective for cutting bone especially in specialist situations such as preparing site for the place of dental implants, several authors report on new advances of the instrumentation [11, 12]. These new designs either resemble their rotary counterparts or are based on more powerful longitudinal ultrasonic vibrations. They offer more precision and better cutting of the bone structure with minimal damage. All studies confirm that ultrasonic cutting needs to be used under medium pressure (around 100 to 150 g), and the type of

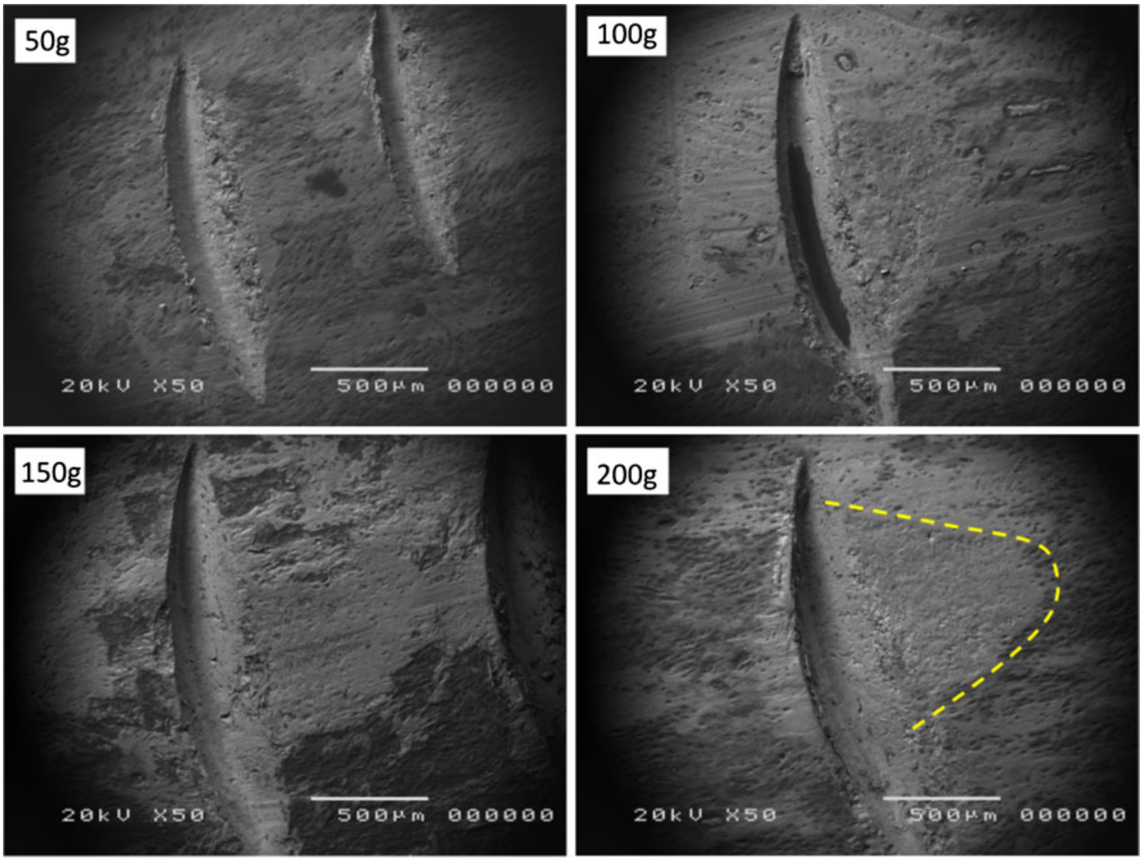

Fig. 5 SEM images for contact loads of 50 to 200 g. For 200 g, the *dotted lines* indicate the defect produced during the initial contact phase of cutting

bone that they cut against should be factored into the cutting process. Although previous work has shown the large range of loads that clinicians apply when using an ultrasonic device [13], there are limitations to our study as clinically, such instruments would not necessarily dwell in one place for several seconds. Further work is required on how ultrasound cutting may be altered as it cuts against different bone structures clinically, and this may require finite element models of the process prior to introduction of new clinical tips [11, 14].

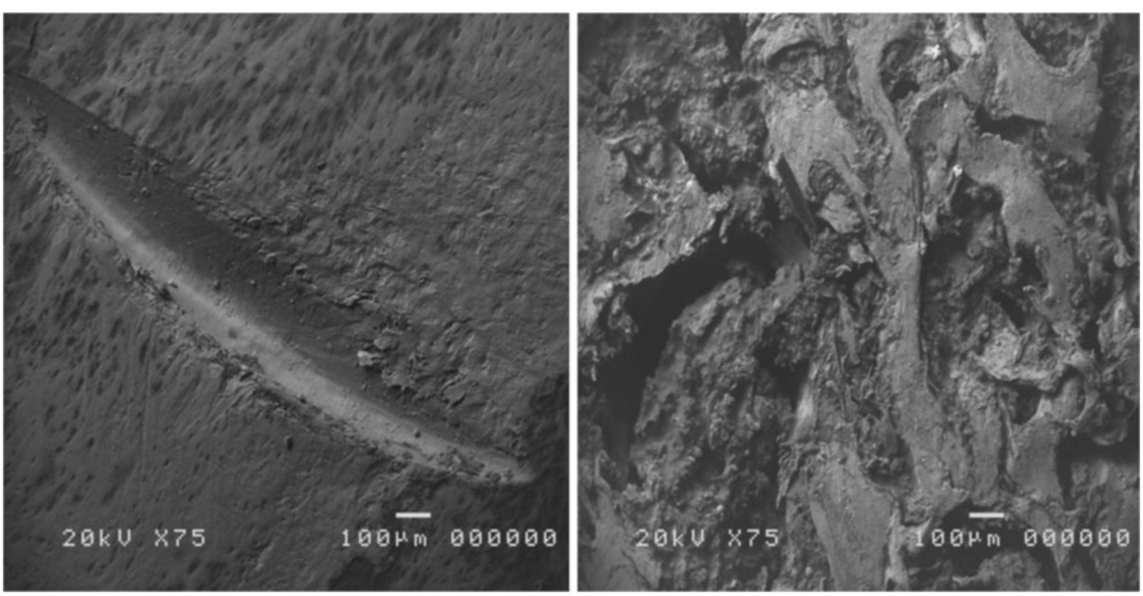

Fig. 6 SEM images for contact loads on the cortical (*left*) and spongious (*right*) bone samples. Images were acquired at 20 kV with ×75 magnification

Conclusion

This study assessed a piezoelectric cutting system with an OP3 style insert tip on porcine bone samples in vitro. A load of 150 g provided the greatest depth of cut when the system was operated in the cortical mode. The structure of the bone was found to be important, with the OP3 tip giving higher cut lengths and widths in the spongious mode compared to the cortical mode. Care must be taken at lower loads to prevent the tip from sliding across the surface of the bone.

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Conflict of interest The authors declare that they have no conflict of interest.

References

1. Salami A, Dellepiane M, Crippa B, Mora R (2010) A new method for osteotomies in oncologic nasal surgery: Piezosurgery. Am J Otolaryngol 31:150–153
2. Arnez J, Papa G, Renzi N, Ramella V, Panizzon N, Tonanetti F (2009) Use of piezoelectric bone scalpel in hand and reconstructive microsurgery. Acta Chir Plast 53:207–214
3. Horton JE, Tarpley TM (1981) Clinical applications of ultrasonic instrumentation the surgical removal of bone. Oral Surg 51:236–241
4. Sortino F, Pedullà E, Masoli V (2008) The piezoelectric and rotatory osteotomy technique in impacted third molar surgery: comparison of postoperative recovery. J Oral Maxillofac Surg 66:2444–2448
5. Lea SC, Walmsley AD (2009) Mechano-physical and biophysical properties of power-driven scalers: driving the future of powered instrument design and evaluation. Periodontol 2000 51:63–78
6. Felver B, King DC, Lea SC, Price GJ, Walmsley AD (2009) Cavitation occurrence around ultrasonic dental scalers. Ultrason Sonochem 16:692–697
7. Parmar D, Mann M, Walmsley AD, Lea SC (2011) Cutting characteristics of ultrasonic surgical instruments. Clin Oral Impl Res 22:1385–1390
8. Lea SC, Landini G, Walmsley AD (2004) Assessing the vibrations of dental ultrasonic scalers. J Sound Vib 271:1113–1120
9. Pavlíková G, Foltán R, Horká M, Hanzelka T, Borunská H, Sedý J (2010) Piezosurgery in oral and maxillofacial surgery. Int J Oral Maxillofac Surg 40:451–457
10. Romeo U, Del Vecchio A, Palaia G, Tenore G, Visca P, Maggiore C (2009) Bone damage induced by different cutting instruments—an in vitro study. Braz Dent J 20:162–168
11. Cardoni A (2010) Enhancing oral implantology with power ultrasonics. IEEE Trans Ultrason Ferro Freq Cont 57:1936–1942
12. Alam K, Mitrofanov AV, Silberschmidt VV (2011) Experimental investigations of forces and torque in conventional and ultrasonically-assisted drilling of cortical bone. Med Eng Phys 33:234–239
13. Khambay BS, Walmsley AD (2000) Investigations into the use of an ultrasonic chisel to cut bone. Part 1: forces applied by clinicians. J Dent 28:31–37
14. Lucas M, MacBeath A, McCulloch E, Cardoni A (2006) A finite element model for ultrasonic cutting. Ultrasonics 44:503–509