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Gauging electrical properties of bone with a bioimpedance-sensing drill

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Abstract

Objective: Multiple surgical procedures require drilling through bone to access underlying anatomy or for device placement. In dental applications, iatrogenic injury caused by accidental perforation of cortical bone into critical anatomy occurs in a significant number of dental implant procedures. Limited real-time sensing methods are available to guide clinicians while drilling through cancellous bone towards cortical interfaces. Early interface detection could prevent accidental damage to underlying anatomic structures. This study aims to demonstrate that electrical impedance spectroscopy can be integrated within a standard surgical drill to distinguish between bone types.

Approach: A custom bearing and insulated drill bit were designed to collect 10 cortical and cancellous impedance measurements from *ex vivo* femurs (2 pigs) and 80 bone measurements from *in situ* femurs (4 pigs). In the *in situ* study, two return electrodes (simulating a lip-clip and shoulder pad) were used to assess impedance differences associated with return electrode geometry.

Main results: Significant *ex vivo* differences ($p < 0.05$) between cortical and cancellous bone occurred in the real resistivity from 0.1 to 100 kHz and in the imaginary resistivity from 0.3 to 10 kHz, with a maximum cortical-to-cancellous impedance ratio of 1.48. Significant *in situ* differences ($p < 0.01$) were observed in both real and imaginary resistivities from 0.1 to 100 kHz, with a maximum impedance ratio of 2.94. AUCs for classifying bone type based on the real resistivity ranged from 0.84 to 0.96 for *ex vivo* bone and 0.98 to 1.0 for *in situ* bone. Mean differences between return electrode geometries were less than 5%.

Significance: The significant cortical-to-cancellous contrast observed indicates this system's potential to provide real-time tissue differentiation during bone drilling procedures.

Keywords

bioimpedance; electrical impedance spectroscopy; bone drilling; bone impedance

1. Introduction

Drilling through bone, to access underlying anatomy or to place a device, is common in multiple surgical fields, including orthopaedics, neurosurgery, otolaryngology, and oral-maxillofacial surgery. As clinicians progress through the bone, special care must be taken to ensure the drill bit does not breach through the outer cortex and impact neighboring structures, such as nerves, vessels, or critical organs. Studies have shown that clinician control over a surgical drill's course is variable; on average, the drill bit will penetrate 6.31 mm beyond the cortex (Clement et al 2012).

In dental applications, implant procedures have been shown to have a 40% rate of iatrogenic injury from bone-drilling complications (Juodzbalsys et al 2011). A dental implant is a titanium screw placed into the alveolar (cancellous) bone of the maxilla or mandible to act as a replacement root for a missing tooth. Implantation involves creating an initial osteotomy in the cancellous bone with a 2.0 mm twist drill to establish depth and orientation, followed by enlargement of the hole using increasing-sized drills until the final screw can be inserted. During this initial osteotomy, the drill bit must remain a minimum of 1–2 mm from several critical structures along the drill bit's trajectory. The inferior alveolar nerve (IAN) runs through the mandibular canal directly beneath the molars and divides into the incisive branch and mental nerve at the mental foramen (Gray 1918). Though within the medullary space, the IAN and incisive branch are surrounded by cortical bone; breaching this layer and contacting nerve can result in mild to permanent neurosensory disturbance (Kim 2011). In the maxilla, most teeth in the back of the mouth lie immediately below the maxillary sinus, also lined with cortical bone. Perforation into this sinus and damage to the Schneiderian membrane lining may contribute to sinus complications including acute and chronic infections (Ardekian et al 2006, Misch and Wang 2008).

To prevent damage to these structures, clinicians have relied on radiographs, preoperative planning, and surgical guides (Almog et al 2001, Nickenig and Eitner 2007, Ramasamy et al 2013). Newer CT-guidance and computer-navigated approaches also have been reported (Casap et al 2004, Brief et al 2005, Gulati et al 2015). While these approaches have reduced complications, multiple inaccuracies still lead to inadvertent local injuries (Benavides et al 2012). Average positioning errors are reported to be 1.39 mm with maximum errors >7 mm at the apex of the osteotomy for state-of-the-art navigation-based systems (D'haese et al 2017); this level of error is close to or exceeds the suggested 1–2 mm safety zone typically specified for placing implants.

Electrical properties of tissue provide the ability to differentiate between various tissue types based on cellular content and architecture. Several reports have demonstrated significant contrast between cancellous and cortical bone, with cancellous bone having a higher conductivity (lower resistivity) than cortical bone (Saha and Williams 1995, Gabriel et al 1996a). Integrating electrical property sensing into bone drilling applications could provide real-time clinician guidance in an otherwise nonvisible area. Bolger *et al* reported a handheld bipolar electrical conductivity sensing device integrated into a standard pedicle perforation tool that can detect pedicle breaches with 98% sensitivity and 99% specificity (Bolger et al 2006, 2007). However, the tip diameters (2.5–4.0 mm) and device lengths are too large for

use in the limited working volume of the mouth (Suess and Schomacher 2016) and other space-constrained procedures. Dai *et al* discuss a custom electrical impedance sensing drill for spinal surgery applications (Dai et al 2014). In addition to the larger drill size (>2 mm diameter), their study introduces a new drilling device, rather than modifying an existing clinical probe, and does not account for impedance changes as a function of drill insertion depth.

Here, we describe an electrical impedance-sensing device integrated onto a standard dental drill that can utilize the impedance properties of bone to potentially detect when a surgical drill bit approaches a cancellous-cortical bone interface. This manuscript describes the device design and characterization, details the tissue volume sensed by the drill bit, and highlights results from *ex vivo* and *in situ* measurements that demonstrate the device's ability to distinguish between bone types. To the best of our knowledge, this represents the first account of electrical properties measured *in situ* from both cortical and cancellous bone.

2. Methods

2.1. Device development

A standard dental implant drill (OsseoSet™ 200 system, Nobel Biocare, Kloten, Switzerland) was equipped with impedance-sensing capability through use of a custom designed drill bit and bearing interfaced to an impedance analyzer (figure 1). A 2 mm twist drill was selected as this diameter bit is used to create the initial osteotomy, which establishes total depth and is thus the only blind drilling procedure. An intracannular bearing was designed to robustly interface the electrical hardware to a rotating cannulated drill bit without decreasing the working volume available to the surgeon. The bearing consists of 30 AWG wire soldered onto a 0.25 mm diameter brass rod shaped identical to the OsseoSet's 'Cannula for internal cooling'. The wire has a J-hook protruding from the distal end that provides direct electrical contact with the inner surface of the cannulated drill bit (figure 1(a)). The portion of the bearing outside of the cannula was insulated to minimize electrical coupling between the bearing and drill surface.

Previous work with bipolar systems report impedance changes as a function of probe insertion depth (Bolger et al 2006, Yun et al 2016). To minimize this depth dependence, a 15 µm thick diamond-like carbon (DLC) coating (IBC Coatings Technologies, Lebanon, IN) was applied to the surface of a standard 2 mm cannulated twist drill (Type 36117, Nobel Biocare, Kloten, Switzerland) everywhere except the distal 2 mm and within the cannula via a Plasma Assisted Chemical Vapor Deposition (PACVD) process. DLC was chosen for its extremely high hardness (4000–9000 HV), high resistivity up to 10⁶ Ω cm, biocompatibility, and established use in other medical devices (Thomson et al 1991, Grill 1999, Honglertkongsakul et al 2010). This insulating coating allows for electrical contact at only two points (the sensing tip and intracannular space in contact with the bearing), localizing impedance sensing to the drill bit tip. The length of the exposed tip is a tradeoff between contact impedance (smaller surface area leads to higher contact impedance) and local sensitivity (larger surface area samples a larger volume of tissue, limiting local sensitivity). Preliminary experiments suggested that an exposed tip of 2 mm served as a compromise between these two tradeoffs and was sufficient to demonstrate proof of concept.

A cable harness (coaxial RG188, 2.3 mm diameter insulated cabling) interfaces the drill bit and bearing, as well as a distal return electrode, to an HP 4284A Precision LCR Meter (Agilent Technologies, Santa Clara, CA), controlled via a laptop computer. A custom MATLAB (MathWorks, Natick, MA) user interface records resistance R and reactance X at 31 discrete frequencies logarithmically spaced between 0.1–100 kHz. While this impedance analyzer can measure up to 1 MHz, 100 kHz was chosen as the maximum frequency to limit the effects of parasitics associated with the ~1 m long cable harness. A 1 mA current is injected between the drill bit and return electrode and complex electrical impedance, $Z = R + jX$, is computed from the ratio of measured voltage to injected current. The current magnitude can be easily scaled at lower frequencies (<1 kHz) to meet IEC60601-1 safety standards (IEC 2005). Standard open (no contact between drill bit tip and return electrode) and short (both electrodes in contact) calibration protocols are used prior to measurement acquisition (Halter 2007).

2.2. Bench-top evaluation

Various materials with known impedances were measured to evaluate the device's accuracy in sensing resistive and reactive loads expected within the bone drilling environment. Impedance spectra of discrete resistors from 1 to 10 k Ω (expected range of bone impedances, normalized to our electrode and sample geometries (Gabriel et al 1996a)), biologically equivalent saline solutions from 0.01 to 1 S m⁻¹ (expected range of bone conductivities), and three soft biological materials were recorded with the drill bit rotating constantly at 2000 rpm (figure 2). As expected, the resistance is proportional to resistive loads for discrete resistors and saline solutions, while the reactance is low (0.1° max for resistor loads and 4° at 100 kHz for the saline loads). In the biological materials, the reactive component is much larger, with phases of up to 33° at 10 kHz for the squash sample.

2.3. Measurement stability

Measurement stability was computed across several frequencies for three possible drill bit states: stationary, rotating, and rotating while moving up and down (to mimic the natural drilling motion surgeons employ). The stability was characterized in 0.01 S m⁻¹ and 0.1 S m⁻¹ saline baths, representative of reported cortical and cancellous bone conductivities, respectively (Gabriel et al 1996a). One hundred measurements of R and X for each state-conductivity combination were acquired at 100 kHz (maximum noise occurs at the highest system frequency). The impedance magnitude was calculated as $|Z| = \sqrt{R^2 + X^2}$ and measurement stability was calculated as $\text{SNR} = 20\log_{10}(\mu/\sigma)$, where μ is the mean and σ is the standard deviation of the 100 impedance magnitude measurements.

2.4. Sensitivity analysis

Treatment planning guidelines recommend a 1–2 mm safety zone between the base of the osteotomy and the mandibular canal or maxillary sinus wall (Kim 2011). Thus, it is important to determine the tissue volume around the drill bit tip contributing to the measured impedance, termed the device sensitivity. Axial and radial sensitivities of the drill bit were empirically determined by translating a drill bit submerged in a saline-filled glass tank away from a high impedance boundary in 0.5 mm increments up to 10 mm (figure 3(a)). A large

10 cm × 15 cm brass return electrode positioned far from the drill bit simulated a patient return electrode.

As the drill bit was translated from the high impedance boundary (glass or air), the percent change in impedance with respect to bulk saline was computed. Computationally, sensitivity was defined as the smallest distance at which $\frac{|Z|_d - |Z|_{\text{saline}}}{|Z|_{\text{saline}}} < 4\%$, where $|Z|_d$ is the impedance magnitude at distance, d , $|Z|_{\text{saline}}$ is the impedance magnitude of the bulk saline solution (calculated with the drill bit placed in saline, halfway between the high impedance boundary and air, where measurements are independent of these boundary impedances), and 4% represents the maximum expected noise as determined through the measurement stability assessment. This distance corresponds to the zone around the drill bit tip that has a measurable impact (i.e. above the noise floor) on the recorded impedance. Sensitivities calculated within the homogenous saline bath are expected to be slightly different than those occurring in inhomogeneous bone. However, we expect the sensitivity to fall between that expected for homogenous cancellous and homogenous cortical bone and that these results would indicate the range of distances over which the drill bit is capable of sensing.

2.5. Bone sample preparation

Bone samples were harvested from pig femurs to isolate sufficiently large regions of individual bone types while minimizing electrical contributions from neighboring tissues. Samples from the diaphysis were chosen for pure cortical bone specimens, as cortical bone is the only bone type present in the femur shaft, while samples from the epiphysis center were chosen for cancellous bone specimens, taking care to cut specimens from above and below the growth plate where only cancellous bone was present. For the *ex vivo* experimental set, femurs from two freshly harvested adult pigs were refrigerated for 2–4 h during transportation, brought to room temperature, and cut into five 1 cm × 1 cm × 3 cm samples per bone type. For the *in situ* experiments, impedance spectra were recorded from the femurs of four ~18 kg pigs less than 1 h after euthanasia to as closely as possible mimic *in vivo* conditions. The femur diaphysis and epiphysis center were exposed in turn to locate 10 regions of pure cortical and pure cancellous bone from each pig.

2.6. Cortical-to-cancellous impedance ratio

Ex vivo and *in situ* impedances of the cancellous and cortical bone specimens prepared above were collected to evaluate whether the two bone types could be differentiated with our device. Two-sample t-tests assuming equal means and the ratio of cortical-to-cancellous resistivity ($\rho_{\text{cortical}}/\rho_{\text{cancellous}}$) were used to assess impedance differences between the bone types. Receiver operating characteristics (ROC) curves and associated areas under the curve (AUC) were computed to evaluate the system's ability to classify bone types. All statistics were performed using MATLAB's Statistics and Machine Learning Toolbox.

For *ex vivo* experiments, previous studies concluded that fluid-saturated bone most closely mimics the *in vivo* bone environment (Kosterich et al 1983, Singh and Saha 1984). Therefore, specimens were submerged in a 0.3 S m⁻¹ saline bath, equivalent to the reported conductivity of soft tissue (Gabriel et al 1996a and 1996b) expected to surround the

jawbone. A standard, uncoated 2 mm twist drill was inserted 3 mm deep into the center of one face of each specimen. A 10 cm × 15 cm brass return electrode was positioned parallel to and 10 cm away from the drill bit. Three frequency spectra were acquired per sample and averaged. A total of five samples of each bone type were probed.

For *in situ* experiments, our custom DLC-coated drill bit was inserted 4 mm deep into 10 regions of pure cortical and pure cancellous bone from each pig. Both a 2 mm diameter stainless steel rod (to simulate an electronic apex locator lip-clip commonly used in dental practice) and a 2" round cloth pad electrode (to simulate a large shoulder-located return electrode) were positioned 10 cm away from the drill bit within the surrounding soft tissue to evaluate the impact of return electrode geometry. Three frequency spectra were recorded per region and averaged for a total of 40 R and X measurements per bone type and return electrode geometry. R and X were converted to the real and imaginary parts of the complex resistivity, ρ_R and ρ_X , respectively, for each drill bit electrode using a saline-based calibration procedure (Halter 2007).

3. Results

3.1. Device characterization

Bench-top evaluation demonstrated that the device can record impedance from discrete resistive loads with >99% accuracy across all frequencies (figure 2(a)) and can distinguish between bone-mimicking saline solutions (figure 2(b)). The resistance ratios, averaged across frequencies, between the solutions were: 8.3 (expected 9.4) for 0.011 S m⁻¹ and 0.101 S m⁻¹, 2.7 (expected 3.0) for 0.101 S m⁻¹ and 0.304 S m⁻¹, 2.9 (expected 3.3) for 0.304 and 1.008 S m⁻¹, and 7.9 (expected 10.0) for 0.101 S m⁻¹ and 1.008 S m⁻¹ solutions. Finally, the device accurately can record varying impedance spectra while drilling through soft biological materials with frequency-dependent impedances (figure 2(c)).

Table 1 presents the 'worst-case scenario' measurement stability values (SNR) at 100 kHz. While measurement stability worsened with the drill in operation (from ~60 dB to ~25 dB), the impedance magnitudes for the two conductivity solutions were sufficiently far apart that their values did not overlap, even with the increased noise level (figure 4). With the drill rotating and moving up/down, the minimum impedance difference between the two solutions was 4.97 kΩ, approximately six times the average impedance of the 0.1 S m⁻¹ solution (0.75 kΩ).

The drill bit sensitivity corresponds to the distance at which the impedance magnitude appears to flatten, indicating when the high impedance boundary at $d = 0$ mm no longer impacts the measured impedance (figure 3(c)). Minimum sensitivities were 1 mm in the distal axial (bottom) direction, 2.5 mm in the proximal axial (top) direction, and 1.5 mm radially. Axial sensitivities exhibit a slight frequency dependence compared to the relatively constant radial sensitivity values (figure 3(b)), due to moderate signal coupling through the DLC coating at higher frequencies; however, the decrease in distal axial sensitivity and increase in proximal axial sensitivity as frequency increased was consistent.

3.2. Cortical-to-cancellous impedance ratio

Significant *ex vivo* impedance differences ($p < 0.05$) between cortical and cancellous bone were observed in ρ_R from 0.1 to 100 kHz and in ρ_X from 0.3 to 10 kHz (figure 5(a)). The corresponding conductivities for cancellous and cortical bone are reported in table 2. The cortical-to-cancellous impedance ratios ranged from 1.28 to 1.48 for ρ_R and 0.93–1.44 for ρ_X (table 3). AUCs for *ex vivo* ρ_R and ρ_X ranged from 0.84 to 0.96 and 0.48–1, respectively.

Significant *in situ* impedance differences ($p < 0.01$) were observed in both ρ_R and ρ_X for all frequencies (figure 5(b)). The measured average cancellous and cortical conductivities (including both return electrode geometries) are reported in table 2. The cortical-to-cancellous impedance ratios ranged from 2.82 to 2.94 for ρ_R and 2.22 to 2.86 for ρ_X (table 3). No significant resistivity differences ($p > 0.05$) were observed between the return electrode geometries in cortical bone for all frequencies. Significant resistivity differences ($p < 0.05$) were noted between return electrode geometries in cancellous bone (except in ρ_X from 12.6 to 63.1 kHz); however, the mean differences were small (~5%) compared to the cortical-to-cancellous contrast (~200%–300%). AUCs for *in situ* ρ_R and ρ_X ranged from 0.98 to 1 and 0.80 to 0.99, respectively.

4. Discussion

A custom intracannular bearing and DLC-coated drill bit were designed to equip a standard dental implant drill with bioimpedance-sensing capabilities. Benchmark testing demonstrated that this device can accurately sense resistive loads, physiologically mimicking saline solutions, and biological materials under standard operation. Further characterization of the tissue volume capable of being sensed revealed that tissue 1 mm axial and 1.5 mm radial to the drill bit tip predominantly influences the impedance being measured. These distances imply that as the drill bit approaches a cancellous-cortical interface, a measurable impedance change is expected at least 1 mm away from the interface. This distance falls within the minimum 1–2 mm safety zone recommended for dental implant placement, suggesting that this system could alert clinicians to an approaching interface.

The different sensitivity distances create an ellipsoidal sensitivity volume around the active electrode area; this non-symmetrically shaped volume could have an important effect when considering the 3D spatial arrangement of other tissues or structures around the implant site. In the mouth, neighboring tooth roots might pose the largest influence as their surfaces have a significantly higher impedance, reported to be 30 times larger than that in bone (Huysmans et al 1996). While most structures will remain outside the small sensitivity distances (i.e. 1.5 mm radial to the drill bit tip), the frequency-dependent nature of the sensitivity (figure 3(b)) would enable select frequencies to be chosen to detect an interface or to locate sensitive anatomy at shorter or greater distances as needed (e.g. 10 kHz has a radial sensitivity of 2 mm).

Significant cortical-to-cancellous contrast was observed with this device for both *ex vivo* and *in situ* scenarios. The maximum ratio was 1.48 (at 0.25 kHz) in the *ex vivo* study and 2.94 (at 0.20 kHz) in the *in situ* study. Furthermore, AUCs (figure 5(c)) suggest an almost perfect

accuracy for classifying bone as cancellous or cortical. The cortical and cancellous bone conductivities measured here are within the range of those reported elsewhere (De Mercato and Garcia-Sanchez 1988, 1992, Gabriel et al 1996a). However, literature reports of bone conductivity acquired from machined *ex vivo* samples vary greatly, depending on the type of animal, sample preparation, and experimental setup (De Mercato and Garcia-Sanchez 1988, 1992, Gabriel et al 1996a). For example, Gabriel *et al* reports cortical and cancellous bone conductivities ranging from 0.006 to 0.03 S m⁻¹ and 0.06 to 0.2 S m⁻¹, respectively (Gabriel et al 1996a). Multiple reports have indicated that bone conductivity measurements are largely influenced by the conductivity of the saline solution surrounding the bone and penetrating its pores (Chakkalakal and Johnson 1981, Kosterich et al 1983, Singh and Saha 1984). Since our *ex vivo* samples were placed in a 0.3 S m⁻¹ saline bath, the conductivities of both bone types were increased proportionally with respect to our *in situ* measurements, which more closely represent *in vivo* conditions. However, even with the influence of saline surrounding the drill bit on the resistivities collected from the *ex vivo* experiment, the differences between the measured resistivities within cortical and cancellous bone were significantly different. This result implies that electrical detection of different tissues in the presence of blood or while using an irrigation solution when drilling is still possible. Future work will determine if we can leverage the impact of the irrigation solution's conductivity on the measured impedance for enhanced interface detection.

Several limitations are associated with this study. First, bipolar systems are more prone to electrode contact impedances than tetrapolar systems (Brown et al 2000). However, with such highly contrasting tissue types, reported as a 2:1 to 20:1 resistivity ratio in literature (Saha and Williams 1995) and found here to be a maximum 2.94:1, electrode artifacts and system noise do not overshadow the signal of interest making impedance sensing in this context a robust strategy for tissue detection. Furthermore, heat generation, either between the drill bit and bone or between the bearing and drill bit, should not significantly impact the signal as bone drilling during the implant procedure is limited to no more than 10 s of operation as the surgeon drills the bone in an up/down motion.

Furthermore, bone measurements were taken in the femur as opposed to the mandible or maxilla. The femur was chosen for its large cortical and cancellous volumes to ensure that our drill bit was sensing only a single tissue type. Additional *in situ* cancellous measurements acquired from the spine and skull of one pig suggest that impedance does *not* vary greatly between bone sites. At 100 Hz, ρ_R was $15.41 \pm 1.14 \Omega\text{m}$ in the femur ($N = 40$), $17.59 \pm 0.68 \Omega\text{m}$ in the pedicle ($N = 4$), $16.27 \pm 1.16 \Omega\text{m}$ in the vertebral body ($N = 10$), and $13.54 \Omega\text{m}$ in the skull ($N = 1$). While these measurements were similar within this pig, slight variations between femur and jaw bones might impact the cortical and cancellous bone impedance magnitude and the level of contrast for dental drilling applications.

The *in situ* study may not accurately represent the impedance of *in vivo* bone. Measurements were performed within 1 h of euthanasia to minimize any change in bone properties. However, variations in bone impedance due to environmental factors (temperature, air exposure, moisture content, conductivity of bathing solution) and biologic factors have been reported (Singh and Saha 1984). After euthanasia, blood perfusion stops, body temperature

decreases, and the bone is exposed to air to access the sampling regions. These processes may change the electrical properties measured *in situ* as compared to *in vivo*.

Finally, the cortical bone surrounding the maxillary sinus and mandibular canal can be extremely thin, up to 1 mm (Yan et al 2015), or of an intermediate density between cancellous and cortical. These anatomic variations may not provide as much tissue contrast as provided in the bulk tissues measured here. However, even with these limitations, the IAN (nerve) and maxillary sinus cavity (air) should provide alternate sources of significant contrast from cancellous bone. Nerve has reported conductivities comparable to cortical bone ($0.01\text{--}0.03\text{ S m}^{-1}$), especially at lower frequencies (Roth 2000, Hasgall et al 2018), and air has extremely low conductivity ($<0.001\text{ S m}^{-1}$).

Despite these limitations, cortical and cancellous impedances obtained with our system indicate that impedance-based bone differentiation is possible. As the drill bit senses impedances 1–2 mm from the drill bit tip, tissue changes within that zone will alter the impedance being measured; this quantifiable change can signify an approaching interface and serve to alert the clinician before injury occurs. Future work will focus on characterizing the interface detection limits of the device, tracking impedance changes as the drill moves through various biological phantoms (bilayer gels) and bone constructs (pig femurs, human cadaver mandibles and maxillas, etc). A FEM model of the drill is under development to test more complicated interface geometries, such as off-angle interface approaches, jaws with varying degrees of bone densities, and the presence of nearby structures that may impact the measured impedances. Various drill bit active electrode areas (i.e. by varying the uncoated tip length and drill size) and irrigation solution conductivities (i.e. applied while drilling and collecting measurements) will be tested to determine the optimal design and experimental considerations for increased interface detection. Algorithms for real-time interface detection based on different electrical property parameters are under development using the high impedance ratio present between cancellous and cortical bone.

5. Conclusion

This paper presents a bioimpedance-sensing surgical drill that can measure and distinguish the electrical properties of bone from 0.1 to 100 kHz. The system, consisting of a novel intracannular bearing and DLC-coated drill bit integrated onto a standard dental drill, recorded significant differences between both *ex vivo* and *in situ* bone, with a maximum cortical-to-cancellous impedance ratio of 2.94. In addition, the drill bit is responsive to high impedance interfaces a minimum of 1 mm axial and 2.5 mm radial away from the drill bit tip. Together, these findings suggest that this system might be used to sense impedance changes as the drill approaches a cancellous-cortical interface. Within the mandible or maxilla, this system has the potential to alert clinicians of this approaching interface and prevent iatrogenic injury associated with breaching the IAN or maxillary sinus. Future work will focus on collecting impedance measurements as the drill moves through various bilayer materials and on developing optimized real-time interface detection algorithms.

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Figure 1.

(a) Proposed system, with the DLC-coated drill bit and intracannular bearing integrated onto a standard dental drill to allow for impedance measurement collection. (b) The custom intracannular bearing (left), consisting of a piece of 30 AWG wire soldered onto a 0.25 mm diameter brass rod, was designed to mimic the shape of the drill's cannula for internal cooling (right). (c) Standard cannulated drill bit coated with 15 µm thick DLC everywhere except the distal 2 mm.

Figure 2.

Experimental setup and corresponding resistance and reactance spectra for bench-top evaluation of our prototype. IA = impedance analyzer, DUT = device under test (i.e. discrete resistors). (a) Discrete resistor testing: experiments confirm that our prototype system is able to accurately gauge resistive loads. (b) Saline solution testing: experiments confirm that our prototype system is able to accurately gauge resistive loads in biologically equivalent solutions. (c) Biological phantom testing: experiments confirm that our prototype system is able to accurately gauge spectrally varying resistive and reactive loads in biologically equivalent materials.

Figure 3.

(a) Experimental setup for the sensitivity analysis of the DLC-coated drill bit. The glass tank was filled 3 cm high with a 0.01 S m^{-1} or 0.1 S m^{-1} saline solution. To measure the axial (top) sensitivity, the air-saline interface was initially aligned with the drill bit's stainless steel tip/coating interface. (b) Axial and radial sensitivities for saline solutions at equivalent cortical and cancellous conductivities from 0.1 to 100 kHz. (c) Example of impedance magnitude as a function of distance from a high impedance boundary at 10 kHz. The black X indicates each sensitivity distance, where impedance changes dropped below 4%.

Figure 4. Impedance measurement stability with different states of drill motion. Data was collected at 100 kHz.

Figure 5.

(a) Mean real and imaginary resistivity ($N=5$) of cancellous and cortical bone samples from day-old *ex vivo* pig femurs in a 0.3 S m^{-1} saline bath as measured with our prototype device.

(b) Mean real and imaginary resistivity ($N=40$) of cancellous and cortical bone samples from *in situ* pig femurs as measured with our prototype device immediately following euthanasia. (c) AUCs for classifying cancellous and cortical bone with our device. AUC ranged from 0.84 to 0.96 based on ρ_R and 0.48 to 1 based on ρ_X in the *ex vivo* experiment and from 0.98 to 1 based on ρ_R and 0.80 to 0.99 based on ρ_X in the *in situ* experiment.

Table 1.

Noise level and SNR of possible drill states at 100 kHz.

State	0.01 S m ⁻¹ (cortical)		0.1 S m ⁻¹ (cancellous)	
	Noise level	SNR	Noise level	SNR
Stationary	0.054%	64.2 dB	0.035%	62.8 dB
Rotating	0.067%	63.9 dB	0.15%	50.3 dB
Rotating up/down	3.83%	27.3 dB	2.79%	29.7 dB

Table 2.Average *ex vivo* and *in situ* bone conductivities.

Frequency (kHz)	<i>Ex vivo</i> conductivity (S m ⁻¹)		<i>In situ</i> conductivity (S m ⁻¹)	
	Cancellous	Cortical	Cancellous	Cortical
0.1	0.101 ± 0.017	0.070 ± 0.003	0.066 ± 0.009	0.029 ± 0.013
1	0.209 ± 0.037	0.149 ± 0.010	0.108 ± 0.020	0.048 ± 0.022
10	0.257 ± 0.044	0.195 ± 0.016	0.123 ± 0.025	0.055 ± 0.026
100	0.271 ± 0.049	0.199 ± 0.017	0.139 ± 0.030	0.060 ± 0.028

Table 3.

Cortical-to-cancellous impedance ratio.

	<i>Ex vivo</i>		<i>In situ</i>	
	Minimum	Maximum	Minimum	Maximum
Real resistivity, ρ_R	1.28 (5.01 kHz)	1.48 (0.25 kHz)	2.82 (6.31 kHz)	2.94 (0.20 kHz)
Imaginary resistivity, ρ_X	0.93 (0.10 kHz)	1.44 (1.58 kHz)	2.22 (100 kHz)	2.86 (2.51 kHz)