Microwave Drilling of Bones

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***Abstract*—This paper presents a feasibility study of drilling in fresh wet bone tissue *in vitro* using the microwave drill method [Jerby et al., *Science* 298 (2002) 587], toward testing its applicability in orthopaedic surgery. The microwave drill uses a near-field focused energy (typically, power under ~200W at 2.45 GHz frequency) in order to penetrate bone in a drilling speed of ~1 mm/s. The effect of microwave drilling on mechanical properties of whole ovine tibial and chicken femoral bones drilled *in vitro* was studied using 3-point-bending strength and fatigue tests. Properties were compared to those of geometrically similar bones that were equivalently drilled using the currently accepted mechanical rotary drilling method. Strength of mid-shaft, elastic moduli and cycles to failure in fatigue were statistically indistinguishable between specimen groups assigned for microwave and mechanical drilling. Carbonized margins around the microwave-drilled hole were ~15% the hole diameter. Optical and scanning electron microscopy studies showed that the microwave drill produces substantially smoother holes in cortical bone than those produced by a mechanical drill. The hot spot produced by the microwave drill has the potential for overcoming two major problems presently associated with mechanical drilling in cortical and trabecular bone during orthopaedic surgeries: formation of debris and rupture of bone vasculature during drilling.**

Index Terms—**Orthopaedic surgery, Mechanical properties, Carbonization, Thermal damage.**

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I. Introduction

DRILLING in bones for orthopaedic and dental purposes has been a common practice for decades, using a variety of drilling bits [1, 2]. Rotary drills used today are efficient; however, they suffer several drawbacks including debris and chips spread resulting in foreign-body-reactions, substantial hematoma at the drilling site, heat generation, difficulties in attaining geometrical accuracy, and wobbling [1, 2]. An alternative, more recent approach employs lasers for bone drilling [3], however, this method may be too costly for large-scale use in the clinical setting. This paper presents a novel approach for drilling of bones, using near-field microwaves [4], and provides the microwave energy characteristics for in vitro drilling in fresh (wet) bones as well as the results of mechanical and histological tests conducted in the microwave-drilled bones.

A. Surgical Drilling in Hard Tissues: Review Up-to-Date

Drilling in bone is a common surgical procedure, which may be required during preparation for insertion of a fixative orthopaedic implant such as nail, screw or wire [5], or before insertion of a bone graft to enhance bone healing [6]. Presently, mechanical rotary drillers are the only type used in the clinical setting. Rotary drilling is performed at a wide range of speeds, from low to moderate (<10,000 rpm) up to ultra-high (>150,000 rpm) [7]. During such rotary drilling, bone debris accumulates around the drilling site (Fig. 1). Although measures are taken to carefully clean bone debris at the preclosure stage of surgery, remaining particles may induce a foreign-body-reaction around the implantation site. Such effects may delay bone healing or interfere with the process of osseointegration of the implant [8]. Rotary drilling also ruptures the vasculature at the drilling site. Substantial hematomas around the sites of rotary drilling were demonstrated using histology of rat femoral diaphysis [5]. Hematoxylin and eosin (H&E) histological staining in the same study also revealed coagulum material in the blood vessel lumens around the drilling site. The coagulum partially or fully occluded the vasculature at the site of drilling [5], imposing a second inhibiting effect on bone healing and osseointegration. Worse still, the ruptured vasculature and

lymph systems provide a portal for infections and for entrance of toxins or wear particles that may originate from the implant [9]. Since the stability of an orthopaedic fixation or a bone graft highly depend on the quality and quantity of the host bone, the above disadvantages of mechanical drilling may lead to longer post-operative recovery periods.

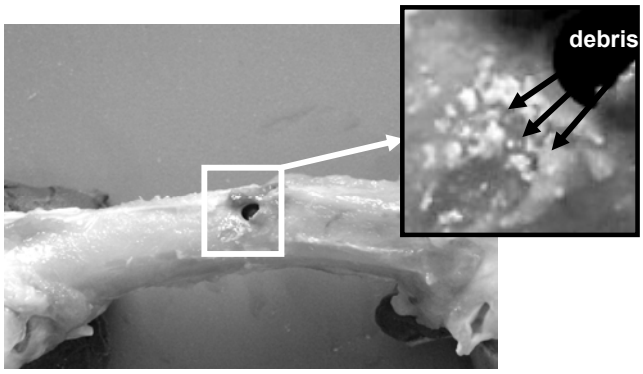

Fig. 1. Debris around the hole after mechanical drilling *in vitro* in cortical bone of chicken femora.

Excessive heat generation in bone tissue during mechanical drilling has been reported, and was attributed mainly to friction during penetration of the rotary drill [7]. The heat generated by the friction between the drilling bit and bone tissue was shown to increase moderately between low and high drilling speeds, and excessively when chips clog the flutes of the drill [7]. Temperatures of 89^{o}C up to 185^{o}C were recorded in distances of 0.5 millimeters from the mechanically-drilled holes [10, 11].

Rotary mechanical drills are difficult to be guided accurately. When touching the smooth bone cortex, rotary drills tend to slide and may dislocate or misalign [2]. Moreover, the diameter of the hole created is larger than the drill bit diameter, due to wobbling effects [1]. After prolonged or repeated use, the dulling of the drill-bit edges increases heat generation and decreases geometrical accuracy. Several attempts for improving the drilling process have been described, all concerned with drill bit geometry [1], drilling mechanism [12] or optimization of parameters (applied force, drilling speed, etc.).

Laser ablation of hard tissues including bones has been compared to rotary drills for a variety of wavelengths and pulse structures [3]. Under proper cooling conditions, the results obtained with laser ablation generally show high accuracy and clean cuts. The main disadvantage in laser drilling is the absent or delayed healing of the ablation site due to photo-acoustic and thermal damage effects [3].

Taken together, the literature above indicates that a new drilling method is needed in orthopaedic surgery. Specifically, such new method should not produce bone debris, should not rupture blood vessels in bone during penetration and should not involve wobbling effects during drilling. Microwave-based drilling can potentially fill these requirements.

B. Microwave-Tissue Interactions

Microwave interactions with biological tissues were studied during the last several years [13,14] but no attempt was made to employ microwave energy for the purpose of drilling in hard tissues. Reported microwave-tissue interactions include heat generation by resistive losses of moving charged-ions and oscillations of charged molecules, and heat transfer by induced movements of charged ions. At temperatures over 50^{o}C, tissues undergo vaporization and carbonization. Higher temperatures may cause desiccation, protein denaturation, coagulation and finally welding and cavitation [13, 14]. It is not yet determined whether RF radiation has additional non-thermal effects on biological tissues, though a recent study indicated the possibility of its carcinogenicity [13]. Tumor ablation by RF radiation has drawn a growing interest recently [15]. The aim of this procedure is to cause coagulation necrosis of cancerous tissue, and thus tumor lysis and ablation. A percutaneous electrode is inserted under imaging guidance (CT, MRI or US) to ablate sub-dermal lesions. The method was also used on bones for ablation of tumors such as osteoid osteoma [16] and metastases [17].

C. The Microwave Drill

A new method for general purpose drilling which employs near-field microwaves was recently introduced by Jerby et al. [4]. The microwave drill device is depicted in Fig. 2a. The coaxial electrode radiates the microwave energy in the near field, thus producing a confined hot spot under the tip of the electrode. The hot spot increases the local dielectric losses of the material in a thermal runaway process [18], allowing the electrode to penetrate deeper into the molten region. This mechanism has no rotating or vibrating parts, and it does not produce debris particles. The heat generated under the drilling tip is anticipated to immediately fuse the bone vasculature crossing the drilling path, thus eliminating hematomas and related coagulum. The microwave drill apparatus (Fig. 2) is considerably less expensive than any laser-based drill.

The objectives of this study were to (i) characterize the microwave parameters (power, drilling speed) for drilling in fresh wet bones *in vitro*, (ii) determine if microwave drilling has degrading effects on mechanical strength and stiffness of fresh whole bones *in vitro* compared with standard mechanical drilling, and (iii) approximate the extent of tissue carbonization at the margins of holes produced by the microwave drill *in vitro*.

II. METHODS

A. Experimental Setup of the Microwave Drill

The experimental setup of the microwave drill is shown in Fig. 2b. The microwave cascade consists a power-controlled magnetron (2.45 GHz) protected by an isolator, a reflectometer for measuring the transmitted and reflected wave power, an E-H tuner for impedance matching, and a

transition from the WR340 rectangular waveguide to the coaxial microwave drill device shown in Fig. 2a. The coaxial structure was cooled by pressurized air (~2 bars). A constant external force of ~20 N was applied axially on the center electrode by means of weights (2Kg), mounted so that their center of mass was aligned with the axis of the electrode.

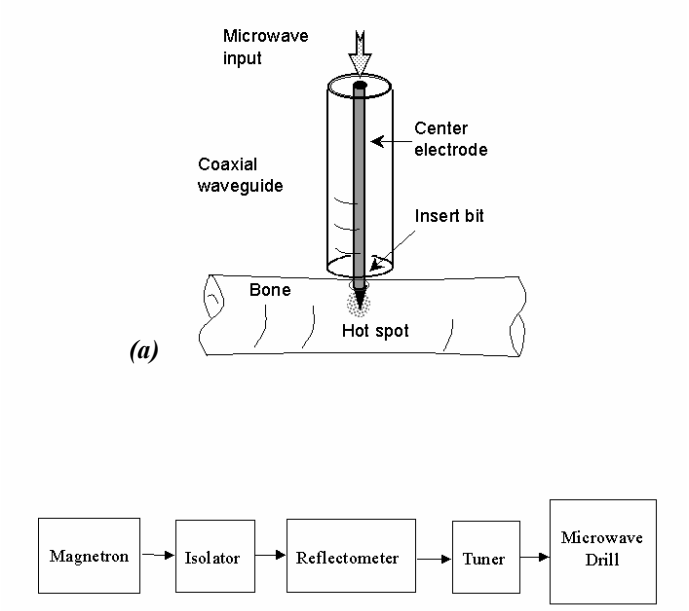

Fig. 2. Scheme of the microwave drill system: (a) the microwave energy is concentrated in a small hot spot in front of the drilling bit, thus enabling its insertion into the bone. (b) block diagram of the microwave drill cascade consisting of a 2.45-GHz magnetron tube protected by an isolator, a reflectometer unit, an impedance-matching tuner (to minimize the reflected power), and the microwave drill device (shown schematically in (a)).

For drilling in bovine trabecular bone (from the proximal tibial diaphysis), an effective power of ~150-200W was required to produce 2.4-mm-diameter holes within 2-5 seconds. For ovine tibias with diaphysial cortical thickness of ~3-5 mm, an effective microwave power of ~200W was sufficient for drilling 2.4-mm-diameter penetrating holes (using an electrode with the same diameter) within less than five seconds. Hence, the effective power needed to drill in bovine trabecular and ovine cortical bones generally resembled. For drilling in chicken femora (with thinner cortex, ~1 mm), a lower effective power of ~60W was required, to produce 1-mm-diameter holes (using a 1mm diameter electrode) within less than 1.8 seconds. Fig. 3 shows for instance microwave drilling results in cortical ovine tibia bone *in vitro*.

A motion detector installed on the microwave drill bit indicated the depth of penetration into the bone. Preliminary drilling experiments were conducted with fresh wet cortical and trabecular bovine bones, to optimize the following parameters for an efficient, low-energy and shortest drilling process: drill-bit depth, impedance matching (minimizing the microwave reflected power), hole geometry, and the scorching radius.

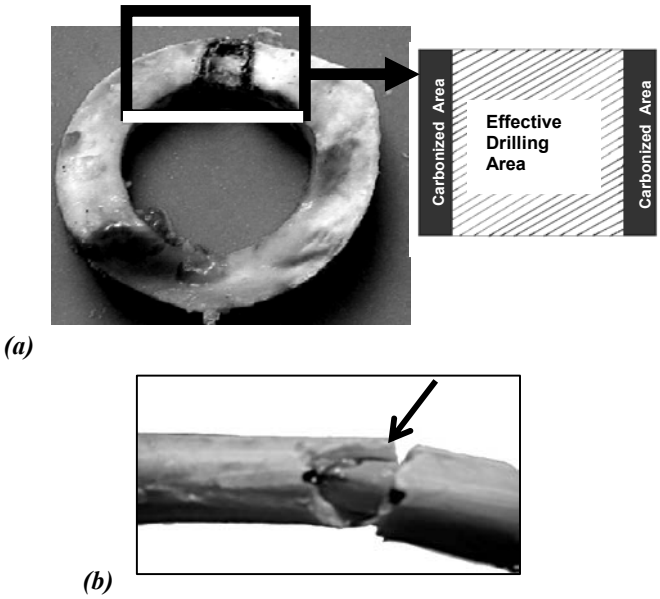

Fig. 3. Microwave drilling in cortical bone *in vitro*: (a) transverse slice in the shaft after microwave drilling (ovine tibia). The carbonized margins around the hole can be identified. The scheme on the right frame demonstrates the method of calculation of the carbonization effect ratio. (b) Microwave drilling showed no visible thermal damage to bone marrow (ovine tibia, with marrow exposed under the fractured bone).

B. Mechanical Property Studies

We determined the feasibility of drilling in bones *in vitro* using the new microwave system, by drilling in bovine cortical and trabecular bone, ovine cortical bone and chicken cortical bone. In order to test the effect of microwave drilling on bone mechanical properties, we drilled holes of the same diameter, in the same location in diaphyses of long bones, using a standard mechanical (rotary) drill and the new microwave drill described above, and compared the mechanical performances of the bones, in terms of bending strength, elastic modulus of the bone shaft following the drilling, and tolerance to fatigue.

B1. Strength and Elastic Modulus

The success of a fracture fixation or insertion of bone graft material is determined mainly by the strength and elastic modulus of the repaired bone. In preparation for mechanical testing of these properties, the soft connective tissues were gently removed from 22 ovine tibia and 18 chicken femur specimens that were obtained fresh from a local butchery. While sheep are a well-accepted and commonly used orthopaedic model, chicken bones are not as widely used, however, in a previous publication [19] we discussed the similarities between mechanical properties of human and chicken cortical bone, particularly, under 3-point bending. These similarities in mechanical properties allow to use chicken femora as an orthopaedic model in the present study. Sheep were 5 months old at the time of sacrifice, and hens were 10 months old (i.e. both had completely ossified bones).

It has been reported that mechanical properties of bone specimens that were frozen at –20ºC (for less than 100 days) and subsequently thawed to room temperature were indistinguishable from properties of fresh bones (for trabecular tissue, cortical tissue and whole bones) [20,21]. Based on these reports, we stored the fresh cleaned tibial and femoral bones frozen at -20ºC until the day of drilling (no longer than 1 month from the time of freezing). Specimens were thawed to room temperature (25ºC) 2 hours before drilling, and were kept moist (using a spray of normal saline) until drilling and during mechanical testing.

In preparation for fixation of long bones (e.g. using screws and plates) drilling is typically performed through cortical bone in the mid-shaft [7]. The experimental design was therefore focused on the potential effects of microwave drilling on cortical bone around that site.

Means and standard deviations of the dimensions of ovine tibias and chicken femora were obtained using a caliper (resolution 0.01 mm) and are provided in Table 1. In order to ensure uniformity of specimens assigned for testing, we allowed standard deviations of less than 10% for bone diameter, less than 5% for bone length, and less than 30% for bone thickness (Table 1). Using two-tailed unpaired *t*-tests we verified that specimens assigned for mechanical drilling (ovine tibia: N=12; chicken femora: N=8) had mid-shaft diameter, cortical thickness, length and cross-sectional area that were statistically indistinguishable from those of specimens assigned for microwave drilling (ovine tibia: N=10; chicken femora: N=10).

TABLE 1
DIMENSIONS OF OVINE TIBIAS AND CHICKEN FEMORA USED FOR DRILLING (MEANS ± STANDARD DEVIATIONS)

Bone dimensions	Ovine tibias	Chicken femora
Major shaft diameter [mm]	20.2 ±1.4	9.21±0.6
Minor shaft diameter [mm]	19.5 ±1.1	8.57 ±0.6
Cortical thickness at the drilling site [mm]	3.9 ±0.7	1.03 ±0.3
Length [mm]	185.7 ±7.6	76.4 ± 3
Cross-sectional area at the drilling site [mm^2]	196.4 ±35.1	25.25 ±5.9

For mechanical drilling, we used a standard rotary drill (MultiPro, Dremel Co.) mounted on a drill press device. The drill press device ensured stability and reproducibility of drilling sites and orientations among bone specimens assigned for mechanical drilling. The rotary drilling rate was 20,000 rpm, corresponding to mid-range drilling rates used clinically [7]. Using either mechanical (rotary) or microwave drills, we created a single penetrating hole in the center of the shaft, with diameter of 2.4+0.1 mm and 1+0.1 mm for the ovine tibias and chicken femora, respectively. In mechanical drills, penetration speed was ~0.7 mm/s. In microwave drills, penetration speed was ~0.8 mm/s for drilling in ovine tibias, and ~0.6 mm/s for drilling in chicken femora. The center of the shaft was identified and marked for each specimen as half the bone length (Table 1).

After drilling, each bone was subjected to a 3-point bending test at a deflection rate of 1 mm/min using an Instron 5544 testing machine (Fig. 4a). The span between the lower supports (*L*, Fig. 4b) was scaled for bone length, and ranged between 13 and 14 cm for ovine tibias, and was set as 4.6 cm for chicken femora. The upper support was pressing against the bone cortex at the side opposed to the location of the hole (Fig. 4b), so that bending-related tensile stresses were applied around the hole. A load cell with maximum capacity of 2 KN was used to measure the applied load (*F*), which was recorded as function of the flexural displacement (*d*) caused by the upper support (Fig. 4b). The failure load (F_{max}) was used to calculate the strength of bone under bending σ [22]:

$$\sigma = F_{\max}\left(\frac{L\,b/2}{4I}\right) \quad (1)$$

where *b* is the minor diameter and *I* is the moment of inertia of the cross-sectional area around the axis of the major diameter (Fig. 4c, Table 1). The slope of the force-displacement curve, $\Delta F/\Delta d$, was further used to calculate the elastic modulus of cortical bone tissue *E* [22]:

$$E = \frac{\Delta F}{\Delta d}\left(\frac{L^3}{48I}\right) \quad (2)$$

We compared each mechanical property (σ, *E*) across groups assigned for mechanical and microwave drilling, separately for ovine tibias and chicken femora, using 2-tailed unpaired *t*-tests. A *p* value less than 0.05 was considered statistically significant in all statistical tests.

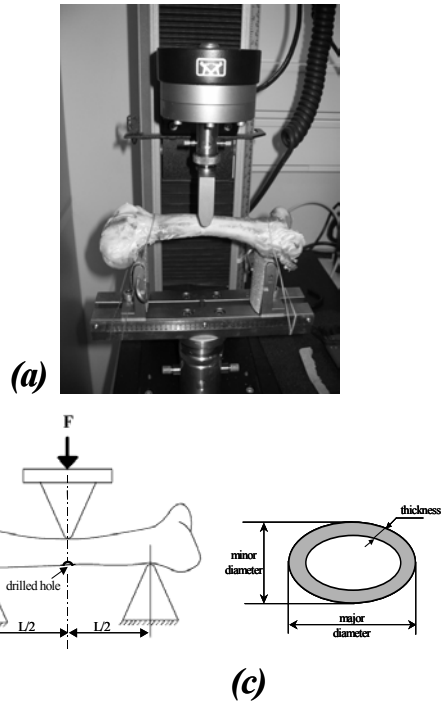

Fig. 4. Mechanical testing of drilled ovine tibias using 3-point bending: (a) the experimental apparatus, (b) Scheme of the experiment showing the lower and upper supporting jigs (triangles). The upper jig applies a flexural force *F* which is balanced by reaction forces *F*/2 at each supporting lower jig. Lower jigs are *L* distance apart. The drilled hole is positioned opposed to the point of application of the flexural force. (c) schematic cross-section through the bone mid-shaft defining the major and minor diameters of the shaft.

B2. Fatigue

Fatigue studies were conducted in chicken femora to simulate the endurance of drilled bones to a more physiological loading scenario (i.e. repetitive, and less than ultimate strength loads). We assigned bones to three experimental groups (i) drilled by microwave (N=10), (ii) drilled mechanically (N=12), (iii) controls, which were not drilled (N=10). One-way ANOVA for each geometrical dimension (Table 1) confirmed that all bone dimensions were insignificant across groups and hence, geometrical uniformity of specimens was verified. Specimen preparation and the drilling process were as described in Section B1.

Chicken femora from all groups were subjected to fatigue in 3-point bending. The load amplitude was set as 220 N, which is ~70% of the load to failure in 3-point bending for the bone geometries considered herein (based on preliminary 3-point bending strength studies, N=12). The loading speed for fatigue was set as 50 mm/min, and the span between the lower supports was again set as 4.6 cm. To determine if microwave drilling had a different effect on cycles to fatigue failure compared with mechanical drilling and controls, we ran a one-way ANOVA.

C. Area of Carbonization

Subsequent to mechanical testing we measured the area of tissue carbonization induced by the hot spot for ovine tibias drilled using the microwave drill. For that purpose, slices of bone from both sides of the microwave-generated hole were cut transversally (Fig. 3a) using an electrically-powered diamond-coated disk saw (MultiPro, Dremel Co.). Measurements of area of carbonization were conducted using image analysis software (SigmaScan Pro, SPSS Inc.) on digital photos of the bone slices (taken at a high-resolution of 2.3 mega pixels). A graph paper was included in the images for calibration of dimensions. We measured the areas of carbonization in the margins of the hole and the projected area of the hole ("effective drilling area") on each slice as shown in Fig. 3a (right frame). We then divided the effective drilling area plus its carbonized margins by the net effective drilling area to obtain a carbonization effect ratio. Ideally, when minimal carbonization occurs, this ratio should approach unity.

D. Optical and Scanning Electron Microscopy

Additional chicken femora were drilled and assigned for optical and scanning electron microscopy (SEM) (microwave-drilled: N=2; mechanically-drilled: N=2). Segments of 9 mm × 9 mm were cut from these bones using a disk saw, so that they contained the hole in their center. Samples were first studied under digital optical microscopy (Axiolab A, Zeiss Co., reflective, magnification ×30). Measurements of spots of carbonization in microwave-drilled specimens were taken using a special ruler for optical microscopy (resolution 100 μm) and a designated micrograph image processing software (SigmaScan Pro).

Second, samples were prepared for SEM studies (JSM840A, Jeol Co., MA, USA). In preparation for SEM, we mounted the bone samples on aluminum discs, using conductive carbon paint. The samples were coated with gold using a sputter coater (SC500, Polaron Co., UK). The acceleration voltage of the SEM was set as 15 KV and the filament current was 3×10^{-10} Amperes.

III. Results

A. Characterization of Microwave Parameters

The variation of effective microwave power with depth of penetration and time during the drilling process is demonstrated in Fig. 5 for ovine cortical (Fig. 5a) and bovine trabecular (Fig. 5b) bone specimens. The effective microwave power for drilling in cortical and trabecular bone components *in vitro* was kept under ~200W and holes were produced after no more than 5 seconds. The effective microwave power varied during the drilling process (Fig. 5) due to the variations in the microwave load-impedance and the consequent reflection from the bone as the drilling process evolved.

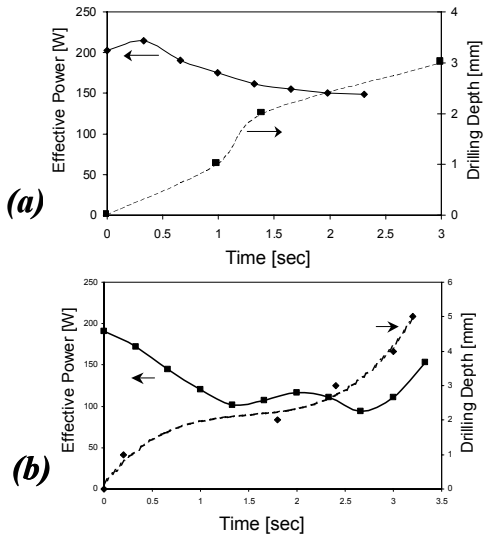

Fig. 5. The microwave effective power (solid line) and drilling depth (dashed line) versus time during microwave drilling through (a) mid-shaft of ovine tibial bone (cortical bone) and (b) bovine trabecular bone.

B. Mechanical Property Studies

B1. Strength and Elastic Modulus

Unpaired 2-tailed *t*-tests showed that mechanical properties of ovine tibias subjected to microwave drilling were statistically indistinguishable from those of tibias drilled mechanically (bending strength ~200 MPa, elastic modulus ~8.5 GPa, Fig 6a,b). Similarly, mechanical properties of chicken femora drilled with microwave or drilled mechanically were indistinguishable (bending strength ~200 MPa, elastic modulus ~5.5 GPa, Fig 6a,b).

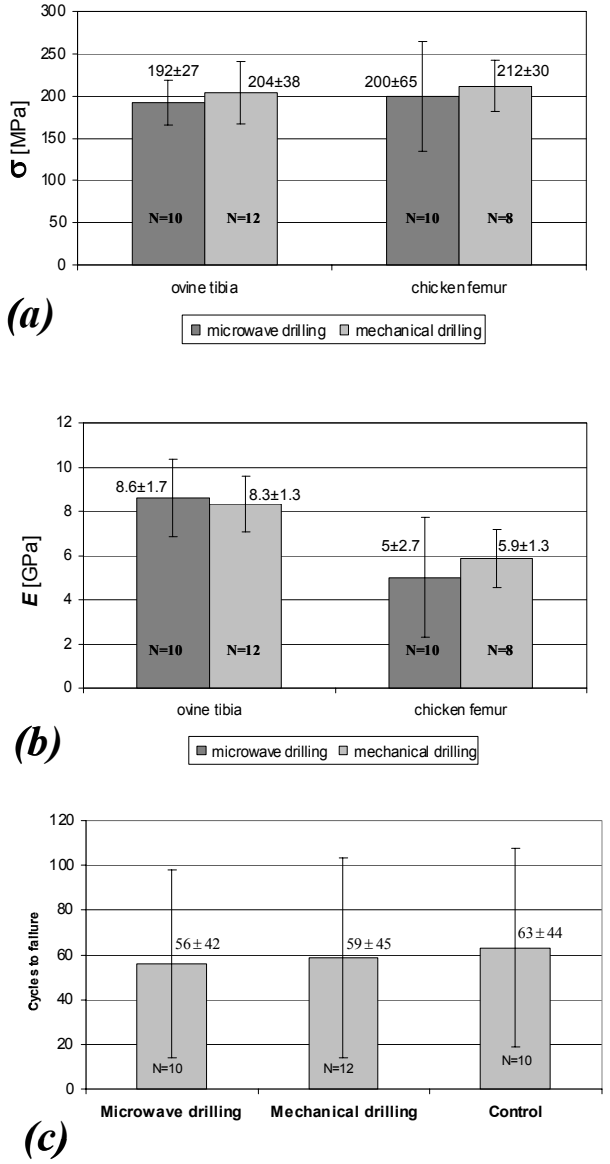

Fig. 6. Comparison of mechanical properties of long bones after drilling: (a) bending strength, (b) elastic modulus and (c) fatigue. Bars indicate means and vertical lines indicate standard deviations.

We conclude that if drilling (microwave or mechanical) reduces mechanical properties of cortical bone *in vitro*, then microwave drilling does not induce a greater deteriorating effect on bone properties than does mechanical drilling.

B2. Fatigue

Using ANOVA, we found that the number of cycles to failure in microwave-drilled chicken femora (mean ± standard deviation 56±42 cycles) was statistically indistinguishable from those drilled mechanically (59±45 cycles). Femora which were not drilled (controls) showed a slightly higher number of cycles to failure (63±44 cycles) (Fig 6c), but without statistically significant difference compared to the drilled femora, very likely because the hole diameter was very small (~1 mm) compared to bone dimensions (length ~76 mm, diameter ~9 mm). We conclude that microwave radiation did not influence the resistance of chicken femora to fatigue.

C. Area of Carbonization

For trabecular bone (bovine), no peripheral surface scorching was apparent, but crosscuts sometime showed slight scorching at the margins of the hole and in bone marrow contained in trabecular spaces. Contrarily, none of the ovine tibias drilled with microwave showed visual evidence of thermal damage to bone marrow contained within the mid-shaft cortex (Fig. 3b). Scorching of ~0.5 mm was visible on the cortical bone surface around the drilled hole (Fig. 3a). The carbonization effect ratio (Fig. 3a) for tibias drilled with the microwave drill was 1.15 ±0.2 (N=10).

D. Optical and Scanning Electron Microscopy

The optical microscopy images of the boundaries of the holes (Fig. 7) show remarkable differences between holes drilled mechanically and with aid of microwave. Specifically, spots of carbonization (size 815-2600 μm^2, measured using image analysis software, Fig. 7c), not seen by visual inspection, can be identified on the hole perimeter in microwave-drilled bone. However, the hole geometry is substantially smoother than in mechanically-drilled bone.

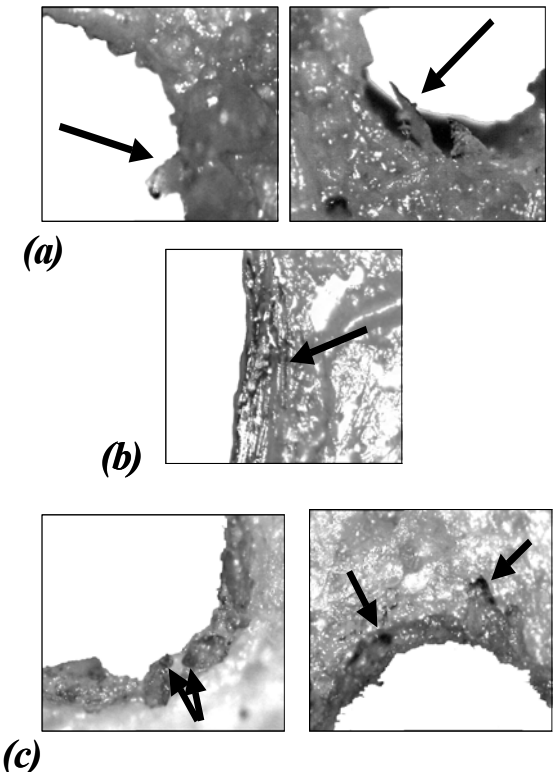

Fig. 7. Digital optical microscopy images (magnification ×30) of hole boundaries in chicken femora drilled-mechanically (a,b) and with microwave (c). For microwave-drilled bone, spots of carbonization (size 815-2600 μm^2) are marked with black arrows around the hole perimeters. Nevertheless, the hole geometry in microwave-drilled samples (c) is substantially smoother than in mechanically-drilled samples (a). Specifically, the mechanically-drilled holes are characterized by sharp fragments that were still attached to the hole surface. Scratches around the mechanically-drilled holes were also identified (b).

The mechanically drilled holes were characterized by sharp fragments that were still attached to the hole surface (Fig. 7a). Around the mechanically-drilled hole, there were some scratches which were likely caused by the initial drilling maneuver to place and stabilize the drill (Fig. 7b).

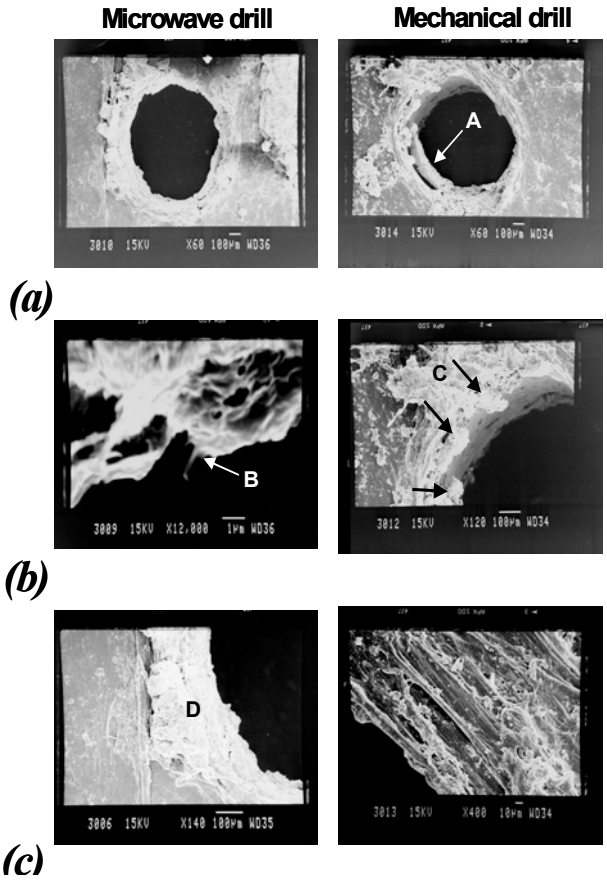

Fig. 8. Scanning electron microscopy (SEM) images of chicken femora drilled with microwave (left column) and mechanically (right column): (a) view of the holes from above, and (b,c) typical imperfections attached to the hole perimeters. Hole geometry is substantially smoother in microwave drilling (a). Typical defects in mechanical drilling are the large fragment of bone partially detached from the hole surface (arrow "A"), sharp bone edges still attached to the contour of the hole (arrow "C"), and scratches ((c), right frame). Typical imperfections in microwave drilling are strut-like elements around the hole perimeter (arrow "B") and small lumps (region "D"), which were apparently formed by melted bone minerals. Magnifications and scales are provided under each SEM frame.

The results of the SEM analyses are consistent with those of optical microscopy, but the high resolution of the SEM detected additional details not seen in optical microscopy (Fig. 8). SEM images of the entire holes demonstrated that hole geometry is substantially smoother in microwave drilling (Fig. 8a,b). Specific examples of defects in mechanical drilling are the large fragment of bone partially detached from the hole surface (marked by arrow "A" in Fig. 8a, right frame), sharp bone edges with characteristic size of 100 μm that were still attached to the contour of the hole after drilling (arrow "C", Fig. 8b, right frame), and scratches with width of about 10 μm around the hole, which were probably caused by drill vibrations during the first contact with bone (Fig. 8c, right frame). SEM images of microwave drilling also showed some typical imperfections likely to be related with the heat generated during the drill. Specifically, we observed strut-like elements around the hole perimeter, with length of less than 1 μm and thickness of less than 0.2 μm (arrow "B", Fig. 8b, left frame), which appear to be hardened fibers of melted bone minerals. A second defect characteristic of microwave drilling was small lumps (dimensions of 100-200 μm, region "D" in Fig. 8c, left frame), apparently of melted bone minerals, which were pushed out of the hole by the microwave electrode.

IV. Discussion

The feasibility of drilling in cortical and trabecular bone tissues using microwave radiation was demonstrated in this *in vitro* study. The hot spot produced by the microwave drill has the potential for overcoming a major problem currently related with mechanical drilling in bone: formation of debris leading to foreign-body-reaction (Figures 1,6,7). When passing through blood vessels in bone, the microwave drill can potentially weld the vessels contacting the hot spot (as opposed to rupture of vessels during mechanical drilling). Hence, the risk for infection may be lower in microwave drilling. The microwave drilling process is relatively quick and is geometrically precise (no mechanical wobbling is involved). Drilling penetration can be monitored and controlled on a timescale in the order of fractions of a second (Fig. 5). Importantly, the microwave drill system is substantially more economical than laser-based drills. The microwave drill technology is versatile and with tailored design, can be considered for a wide range of clinical applications. These may include insertions of orthopaedic pins, nails and screws into bones, wire fixation, neurosurgery, dentistry, cranial surgeries, open-chest cardiac surgeries, and any other procedure requiring accurate, carefully controlled bone drilling.

In vitro drilling in bone provided several encouraging results: (i) microwave drilling did not degrade the mechanical properties of bone in vitro more than rotary drilling did and (ii) carbonization at the margins of the hole in cortical bone was relatively small (~15% of the effective hole diameter) and (iii) holes produced by microwave drilling were substantially smoother than those produced mechanically, and no partially attached bone fragments were observed under microscopy (optical or SEM) in microwave-drilled holes.

We assume that carbonized bone *in vitro* represents the minimal volume of damaged bone tissue around the microwave drilling site. We cannot rule out further tissue damage in non-carbonized regions *in vivo* in bones drilled with microwave, e.g. due to thermal and microwave radiation effects, and this should be quantified in *in vivo* (animal) studies. However, it may be possible to reduce the carbonization margins further before proceeding to *in vivo* studies, using adaptive control on the microwave power, by a

liquid or air cooling mechanism. Basic studies on thermal properties and thermal tolerance of bone tissue under microwave radiation can also be beneficial in order to minimize the carbonized margins of the hole.

Some limitations in this study should be recognized. First, *in vitro* bone specimens, rather than living bones, were used for the rotary and microwave drilling tests. Although bone specimens were obtained fresh and kept moist with a saline spray during testing to closely represent the living bone, the important *in vivo* effects such as viability of osteocytes, tissue necrosis, and hyperemia could not be analyzed at this stage and are awaiting further studies. The interaction of microwave heating with fluid flow in bone vessels is similarly not considered *in vitro*. Second, a drill press was used during the rotary drilling experiments rather than a hand-held orthopaedic surgical drill. To replicate clinical conditions as closely as possible, a drill speed of 20,000 rpm, representative of actual orthopaedic drills was used for all rotary drilling maneuvers [7]. The authors of this paper believe that the advantages of site reproducibility and stability provided by the drill press apparatus were required for the present experimental design, which was aimed at comparing "ideal" rotary and microwave drills.

Safety of the patient undergoing surgery and of the staff who operates the drill is a critical issue to consider at this stage of research, as microwave radiation may be hazardous. A tailored design of the microwave drill and its screening for each specific surgical application is expected to significantly reduce the radiation exposure of both the patient and staff to a permitted minimum. It should be noted that existing RF ablation procedures [e.g. 16] use RF power of up to 150W for about 15 minutes (~900 seconds), much longer than needed for microwave drilling in bones. Operated in a similar power level, microwave drilling through cortical bone only lasts 2-3 seconds (Fig. 5). Nevertheless, research efforts are required to determine whether microwave drilling induces a risk for carcinogenic effects, and how such risk can be minimized.

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