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On chip formation mechanism in orthogonal cutting of bone

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A B S T R A C T

Bone cutting is an important procedure in most surgery operations; therefore, understanding the mechanism by which the chip forms is important to design tools for optimising the surgery process. However, only very few studies to address this issue exist. In this paper, a transition from shear cutting, shear-crack cutting and fracture cutting modes with the increase of uncut chip thickness in orthogonal bone cutting was presented. To address these phenomena, a fracture mechanics based cutting model was proposed for the chip formation and cutting process. The influence of the anisotropy of bone to the orthogonal cutting was studied in reference to chip formation and thus, considered in the model to explain the chip morphologies in different cutting directions relative to the bone fibres (i.e. osteon). The experimental result shows to be consistent well with the proposed model yielding a maximum error of predicted transition uncut chip thickness of 10%. On the other hand, analysis of surface morphology revealed that significant differences exist in material damage mode related to the cutting mode and direction. Moreover, the proposed bone orthogonal cutting mechanism was also validated by the cutting force analysis from both static and dynamic components points of view. Based on this study, a maximum uncut chip thickness in a unique anisotropic material as bone to guide the selection of more advanced cutting process (e.g. drilling, milling and sawing) was proposed.

1. Introduction

Bone cutting is a challenging considering possible tool breakage, structural damage on the living workpiece and thermal necrosis that can accompany the surgical interventions. At the same time, the level of “intimate” contact between the implant and bone has a key contribution to the recovery of the patient after operations [1]. In this case, ensuring of surface integrity and cutting precision is of importance to the success of the bone surgery interventions. As such, before performing those complicated surgical procedures such as orthopaedics and total joint replacement, the study of bone orthogonal cutting is of assistance in supporting the selection of cutting conditions and tooling for improving the quality and safety of these interventions.

Compared to cutting of isotropic materials, the bone surgery deals with a living workpiece structure that is composed of two main tissues with different mechanical properties: the cortical bone in outer layer and the soft tissue (cancellous bone) filling the inner region of cortical layer [2]. Cortical bone with higher density, sustaining elevated mechanical loads, represents the mainly cutting part in any surgery operations. Similar to ceramic matrix fibre reinforced composite, this anisotropic part mainly consists of osteons (fibre) and interstitial lamellae (matrix) in structure and hydroxyapatite in composition, as shown in Fig. 1. Osteons (Haversian system) are roughly cylindrical structures which are typically 3–5 mm in length and ranging between 50 and 300 μm in diameter [3]. Haversian canal which transports blood exists in the centre of osteons. Interstitial lamellae matrix surrounds the osteons and has a lower toughness. Between the interstitial lamellae matrices and osteons there exists a thin layer (1–5 μm) called cement line which is a weak zone (interface) that is prone to crack initiation and propagation around the osteons [4]. In recent years, research interest has been paid on bone cutting related to the surgery application, especially on the hard part of cortical layer [5].

With this anisotropic structure, the bone cutting procedure is strict as some cutting malfunctions could injure the bone cell from physical or histology point of view. Thus, bone cutting calls for specialist cutting technologies/knowledge to assist surgical interventions. In this respect, it has been reported that bone cutting can result in near-surface damage while this is adverse to the post-surgical recovery rate [1]. Further, improving the bone cutting accuracy is important for the implant positioning [6]. For example, it has been reported that in conventional total knee arthroplasty operation the bone surface flatness ranges from 0.16 to 0.42 mm, which reduces the advantage in an accurate positioning of implant and therefore, higher precision in cutting is needed [7]. However,

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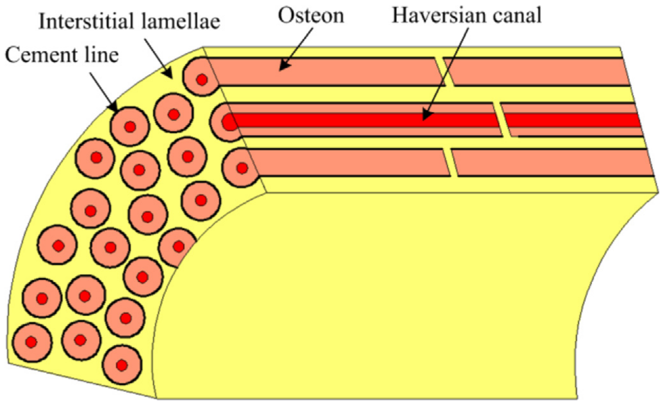

Fig. 1. Schematic of cortical bone structure.

the cracks and/or surface damages can easily develop during cutting since bone is a brittle material [8,9] while these cutting defects could break the Haversian system and influence the regrowth of bone after surgical interventions.

In this respect, studying the cutting and chip formation mechanism of bone machining is of high importance. While most researches concentrate on the postoperative tissue damage [10,11], very few focus on the bone cutting mechanism/chip formation. The first report on this topic was done by Jacobs et al. [12] in orthogonal cutting of bones in 1974. They investigated the cutting mechanics within three different feed orientations related to the osteon direction and found continuous chips fractured with osteons sliding over each other. Wiggins and Malkin [13] made different observations that contradict with research of Jacobs; they suggested the discontinuous chip occurs by a series of discrete fractures. Coinciding with the view of Wiggins and Malkin, Krause [10] proposed that the chip could be considered a continuous structure composed of discrete segments. Christopher et al. [14] and Sugita [15] found that the chip is continuous at small cutting depths while discrete fracture occurs with the increase of cutting depth.

It could be commented that the different opinions on the chip formation mechanisms reported above could be caused by lack of studies considering the relationship between the cutting depth and the chip formation mechanism. The cutting depth (>200 μm) on which Wiggins and Malkin reported in reference to chip formation mechanism could be considered much bigger than the values used in bone surgery when employing drilling [16,17], milling [18] and grinding [19,20]; this could also lead to undesirable outcomes such as high cutting forces and temperatures which can cause cellular necrosis [21]. In reference to Jacobs' [12] experiment, the cutting depth was 12–48 μm while Krause [10] held the constant cutting depth for all tests at 70 μm; thus, their observations were limited in their coverage of the entire phenomenological aspects occurring in bone cutting and, as a consequence, they led to different observed outcomes. On the other hand, it has been shown that the crack behaviour of bone is not constant but related to the loading modes and the osteon orientation [22]. Therefore, it is evident that the chip formation mechanism changes with a change in uncut chip thickness as well as cutting direction. However, these effects are not adequately understood since there is still no proper description on how the chip formation is related to its thickness and anisotropic properties.

In its attempt to cast more light into these aspects, this paper aims to propose a mechanism for orthogonal bone cutting. The chip morphology related to the depth of cut and cutting directions relative to osteon direction was studied. A cutting model which is based on three mechanisms (i.e. shearing, shearing-cracking, fracture) was proposed for chip formation when bone machining. The uncut chip thickness transition of these three cutting modes can be predicted by this model. Cutting forces and surface damages were also investigated and analysed to validate the proposed cutting mechanism. Based on this chip formation mechanism, a maximum cutting depth is proposed for guiding the bone cutting process (e.g. drilling, milling and sawing).

2. Characteristics of bone chip in orthogonal cutting

To study the mechanism of chip formation in bone orthogonal cutting, a series of experiments were first performed to observe the chip formation phenomena. Further, based on these observations the cutting model would be established to explain the cutting mechanism. Finally, model validation was attained through a second series of experiments.

2.1. Specimens preparation and experimental works

The workpiece used in this study was acquired from mid-diaphysis of bovine femurs since the distribution of constituents is very similar to the human bone [23]. The bones were kept in a deep frozen state in physiological saline environment to maintain its mechanical properties with minimal change. Before cutting experiments, the bovine bone specimens were thawed at room temperature for 2 h and pre-cut into 10(length) $\times$ 1(width) $\times$ 4(height) mm^3 pieces in three orientations, i.e. perpendicular, parallel and transverse, to the bone long axis direction, as shown in Fig. 2. The workpiece width (1 mm) was chosen to be smaller than the one of tool cutting edge (2 mm), thus meeting the plain strain conditions of the Merchant analysis for orthogonal cutting [24]. All samples were polished by a 1200-grit emery paper under water condition to obtain smooth surfaces to allow their micrographical analysis. To keep the bone in fresh condition, the experiments were performed within 1 h after being prepared; dry cutting conditions were used throughout the entire orthogonal cutting testing programme.

Based on the relationship of tool feeding direction and osteon orientation, the cutting direction can be divided into three different cases: parallel direction, across direction and transverse direction, as shown in Fig. 3 [25]. These three cutting directions correspond with bone sampling directions as presented in Fig. 2, where the long axes of bone and osteon are in the same direction. The cutting procedure was performed on each red marked surface (Fig. 2) for each cutting direction mode.

An in-house developed 4-axis miniature machine tool [26] was adapted for performing the bone orthogonal cutting tests using a solid carbide cutting edge with rake angle $\gamma = 8°$, clearance angle $\alpha = 8°$ and edge radius $r = 1$ μm, as shown in Fig. 4. The resolution of the machine tool stages is 1 nm with a repeatability of 0.1 μm. Further information about the machine tool setup can be referred to [3] and [26]. The uncut chip thickness was varied from 5 μm and 10 μm to 200 μm (in steps of 10 μm) while keeping constant the cutting speed at a value (33 mm/min) on which the temperature is under the clinically recommended (50 °C) while avoiding the use of cutting fluid thus allowing the observation of chip formation in cutting. A fibre-optic digital microscope (VHX3D Keyence) was set very close to the cutting zone so that chip formation process can be observed during cutting. For the cutting forces measurement a 3-component miniature dynamometer (Kistler 9317B – response frequency <5 kHz) was employed at an acquisition rate of 10 kHz. After machining experiments, the cut surfaces were analysed by Scanning Electron Microscopy (Philips XL30 ESEM-FEG) to observe the cutting damage on the bone material. The cut surface roughness was also measured by surface scanner (Bruker ContourGT-K).